Modular Wearables for Vibrotactile In-Situ Augmented Feedback

Jan Stefan Anlauff

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ABSTRACT

Modular Haptic Wearables for In-Situ Human Performance Augmentation

In this thesis, we present vibrotactile wearables for balance rehabilitation. The tactile, or sense of touch, may be uniquely suited for human-computer interaction in situations where the visual or auditory pathways are unavailable or unsuitable. Vibrotactile wearables, body-worn interfaces using vibration to communicate with the user, can be used ubiquitously and promise to have many applications, for example, in balance rehabilitation research.

Good balance and mobility skills help to prevent falls that can result in serious injury. Balance deficits, for example after traumatic brain injury such as a stroke, are usually addressed in physical and occupational therapy with a set of repetitive exercises adapted to the patient's typical activities. We can improve motor learning and rehabilitation outcomes by providing augmented feedback (AF), external feedback that augments the task-specific, intrinsic feedback during an activity. Vibrotactile feedback is especially suitable for this, as it can guide focus to task-relevant areas of the body, as for example with trunk tilt feedback on the hip during balance excercises.

So far, little research has explored the application of vibrotactile AF for balance rehabilitation outside of clinical environments. Ideally, such augmented feedback aids would be available at home, not require stationary equipment, and be affordable, to support large-scale individualized ongoing, extensive practice for effective treatment in absence of a therapist. Our goal is to bring the state of the art in sensor and actuator technology and the development tools available to creators of such wearables. As balance is the ability to keep the center of mass, located close to the hip, over the base of support, usually the feet, our focus here are augmented feedback wearables in the form of belts and shoes. We focus on systems that combine multiple channels of sensors and actuators supporting physiotherapy excercises for balance rehabilitation. Our contributions are three-fold:

1. We show an augmented feedback belt that provides richer vibrotactile feedback during balance excercises. (jan: *work-in-progress*)
2. We present shoes that can provide stimulation with tactile multi-actuator patterns even during gait and can provide an alternative to auditory feedback for rotatory control inputs.
3. We packaged the tools necessary to create such wearables in an open modular toolkit, enabling rapid prototyping even for non-engineers.

1) Augmented Feedback Belt

Belts providing vibrotactile AF for balance training have demonstrated improved rehabilitation outcomes. Our vibrotactile belt system leverages novel actuators for more expressive feedback incorporating multiple feedback variables, mapping a motion/orientation sensor to four actuators placed in the four cardinal directions on the hip. We present two belt systems for different types of recent vibrotactile actuators. One design is for compact, energy efficient low-cost systems, the other for providing stronger, richer stimuli. A study on healthy participants serves as proof-of-principle for later longitudinal evaluations on populations with decreased balance ability, such as stroke survivors and the elderly.

2) Vibrotactile Shoes

Balance and locomotion are also supported by tactile cues perceived through the sole of our feet, which makes a natural place to provide haptic stimulation. Feet also present a suitable extremity for human-machine interaction when the hands are occupied, as they are fairly dexterous; we commonly use them for control, as with pedals on cars and medical devices. Suprisingly little research has investigated vibrotactile footwear though, especially with regards to untethered multi-actuator systems that can be used during gait. The forces involved in gait present unique challenges for robust and effective actuator integration in vibrotactile shoes. Here, we present instrumented shoes that can provide vibrotactile stimuli on four anatomically motivate locations on the foot sole. We validated our actuator integration design in two studies where the vibrotactile shoe was used as a human-computer interface. The first demonstrated user's ability to distinguish six vibrotactile icons even during gait, the second showed that vibrotactile feedback can serve as an alternative to auditory for a foot-based angular menu selection. (jan: *has no match for belt, structure-wise?*)

3) Modular Prototyping Toolkit

Effective sensor and actuator coupling, robustness, and comfortability are some of the challenges that make developing good wearables most often an iterative, interdisciplinary team effort. Through building shoes and belts, we have developed building blocks for the creation of haptic wearables that incorporate state-of-the-art sensing and feedback modalities. With different applications scenarios and rapidly evolving technology stacks, our open modular system can remain up-to-date longer and facilitates adding, removing or repairing components. Augmented feedback for balance training and novel human-computer interfaces serve only as examplary applications for our toolkit. We describe its use as part of a brain-computer interface for minimally responsive patients to illustrate its use in other realms.

Conclusion

In times of increasing load on health systems, decentral approaches may improve patient adherence and rehabilitation speed by supporting training in absence of therapists. However, applied in-situ research and development is needed to bridge the gap between clinical research and solutions deployed to the homes of patients. Here, we present tools for rapid prototyping of state-of-the-art sensor/actor wearables and studies evaluating these tools. We intend to make technology more accessible to facilitate collaboration and closely involving domain experts and practitioners, enabling functional real-world prototyping to gain the understanding that may bring research from laboratories into the homes of patients.

Contents

1 Introduction (I 80%, S 20%)

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words)

Research Questions

Attempts:

- novel vibrotactile balance feedback mapping/stimuli
- spatial vibrotactile wearables
- What are suitable encodings for vibr-tactile balance feedback that leverage the improved control of
- What are the effects of can the improved control of vibrotactile actuators Wearable directional balance feedback

Words for new haptics:

- wide-band
- improved control
- HD-Haptics
- voice-voil
- higher temporal resolution
- spectral?

Thesis Outline

Sub-questions / Chapters:

- Haptic wearable studies:
  - **Belt**: Waist/Center of mass
    - *Does a pulse-based feedback encoding scheme lead to decreased sway in a balance (or hip tilt targeting) task compared to a traditional ERM intensity-based feedback?*

    * How can we create wearable tools to implement this established type of wearable with state-of-the art voice-coil haptics → Reproduce and improve construction
  - **Insoles**: Feet/Contact point during stance/gait.
    * New type of wearable, novel design of insole with tactors placed unterneath anatomical landmarks of foot: Using spatio-temporal patterns (tactons) as example use case to validate our integration, how can we integrate LRA tactors in an insole so that
    * - it is mechanically robust enough for laboratory studies on tacton delivery during gait?
    * - it is coupling tactors well to the foot and decouples them effectively from each other? How well can users distinguish tactons on such an insole, during stance and gait?
    * Using the tactors to provide feedback on foot rotation, how well does vibrotactile feedback on all/one directional tactor support users in a non-visual menu-selection task, compared to auditory and no feedback?
- How can we, as experienced hapticians, **facilitate prototyping and teaching** of such wearable feedback systems, both for ourselves and for newcomers to the field, creating a reuseable toolkit that extends on exisiting modular development aids and minimizes re-invention?

2 Background and State of the Art (I 75, S 25%)

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words)

> Neuroplasticity simply refers to the fact that the brain is the organ that is built to change in reponse to experience more than any other. Behavioral interventions can produce more specific biological changes than any currently intervention known biological method today.
>
> — Richard S. Davidson

2.1 Sense of Touch

(ToDo: *this needs to be significantly expanded and provide a basis for the subsequent chapters*) The sense of touch, or tactile sense, enables us to perceive the surrounding world through skin (cutaneous) contact, while the proprioceptive or kinesthetic sense lets one perceive the positions, movements, and forces of one's own body [1]. The term, "haptic" encompasses both. In the following, we review the perceptual basis and the state of the art in tactile actuation.

2.1.1 Physiological Basis

(ToDo: *[1] good overview*)

Plantar cutaneous tactile sensation, that is the sense of touch in the foot, is provided by mechanoreceptors similar to the ones in the hand and other glaborous (non-hairy) skin. Spatial tactile resolution is well studied, and varies widely across the body. Together with the head, the hand and feet are the most sensitive to touch.[1] Figure **??** shows their distribution and receptive fields. Fast Acting I (FAI) receptors have a median receptive field size of 6-334 mm^2 and respond to vibrations in the range of 5-40 Hz [3], whereas FA II receptors with a receptive field of 10-1000 mm^2 respond to a frequency range of 40-800 Hz [4]. For tactile notification devices, the Pacinian corpuscles are the most commonly addressed mechanoreceptors.

Plantar sensation is important for human postural control [5] in locomotion. This sensation has been employed in floor-based vibrating mechanisms to

[1] One measure for this is the two-point discrimination threshold, i.e., the ability to distinguish between two points [2].

simulate ground textures [6] and to aid patients with balance and gait deficits [7], using textured or vibrating insoles [8], [9].

Sensory transduction is not a linear, temporally or spatially independent process. Rather, intensity thresholds depend on stimulus type, frequency of stimulation, location on the body [10], and activity. Integration of the information from different receptors is the source of complex sensory phenomena. Understanding this integration process remains an area of active research [4], [11], [12]. Locally, sensory thresholds rise under sustained stimulation, which for example allows ignoring the sensation of an object held against the skin. This "sensory adaptation" begins after only a few seconds of sustained stimulation and is important to account for in applications involving a continuous coupling between user and the stimulating device [6].

It also has been used to desensitize certain mechanoreceptor types prior to experimentation targeting only a select type, e.g., [13](Add Citation: *extend*).

Haptic perception is mediated by integration of multiple types of mechanoreceptors in the skin, but also in muscles and joints. Here, we are concerned with vibrotactile stimuli perceived by cutaneous mechanoreceptors in the skin. Fig. **??** shows a cut through the outer skin levels of hairy skin. These sensory neurons respond to mechanical stimuli such as pressure and vibration and can be categorized by the kind of sensation they primarily perceive, their rate of adaption, and receptive field, i.e., the physical region in which they respond.

Distribution of receptors depends on body location (Add Citation: *this*) and skin type (glaborous, or hairless, and hairy). The fingertips exhibit the highest concentration of tactile receptors, and can be considered analogous to the fovea on the retina for vision [14]. Vibration sensitivity in the fingertips, among other functions, serves perception and discrimination surface texture [15].

The human skin is a highly complex sensory network. Many millions of receptors assure, together with the central nervous system, a permanent and livelong sensitivity. It not only protects the organism, but is necessary basis for many aspects of every day life.

Receptor density varies with body area, especially close to cavities as possible weak points it is very high. In the same fashion, the manipulative areas of the extremities, which have a frequent contact with the environment, are equipped with a higher receptor density. These areas on the palm of the hand, and the foot sole, have “glaborous”, or smooth/hairless, skin. The human skin has total area of 1.7-2m2, weighs 10-14kg, is 1.5-4mm thick, and has a characteristic layering.

Stratum ganulosum: dense network of free nerve endings : 1-3um, pain, chemical substances, very light touch, no special construction, up to 100 free nerve endings per sqare millimeter, i.e., 200 million total, 100 million in the rest of connective tissue.

Stratum papillare: Tactile or *Meissner's* corpuscles ca. 50x120 um, up to 24/mm2; decreasing to 6 in older humans [16], 60 mio total, pressure sensitive, rapid adapting up to 30 Hz/optimal 20-30[16], e.g., for sensing skin deformation, no response to constant deformation.

Stratum basale, lowest layer of Epidermis: *Merkel* nerve endings 50/mm2, 100 mio total, pressure sensitive, both longer lasting and changing, up to 3 Hz.

Stratum reticulare, lower Dermis: Bulbous or *Ruffini* corpuscles 50-120 um, three cylinders with different spatial orientation, anchored in surrounding skin cells. slowly adapting, Senses deeper, continued skin stretch. Also in various connective tissues, ligaments and joint capsules. In combination with muscle spindles they are important for proprioception.

Deep subcutis: Lamellar or *Pacinian* corpuscles up to 200x500 um, light and strong sensations, oblong shape, up to 60 lamellae layers. 10-1000 Hz, optimally between 10-100 Hz, nanometer displacements can elicit response, 10-20nm between 150-200 Hz. for 10 Hz, 14um, 20 Hz, 5.6 um.

Muscle spindles receptors in mucles, 65k spindles in 400 muscle groups, more in smaller than in larger groups. 2-10mm long spindle-shaped weavings of nervefibers and endings meshed with muscle fibers. signals of all spindles used by brain to detect how strongly muscle group is contracted.

GOLGI TENDON ORGANS 1.6mm long, 120um wide, similarly integrated as muscle spindles. at junction of tendon bundles to muscle fibers, sometime to their fascia. give important information on tendon elongation during contraction of muscle. [16]: Functionally, GTOs monitor the tension devel-oped by the muscle and are able to elicit protec-tive spinal reflexes via inhibitory synapses with the corresponding motor neurons 5, 6.

HAIRS part of skin of all primates, one or more horn fibers produced in folicles, slightly slanted, on each a muscle. each folicle is surrounded by ca. 50 tactile receptors, thus providing mechanical protection rather than thermical. very sensitive and can sense earlier than internal receptors, e.g., when insects land.

Tactile sensitivity is commonly expressed through measured thresholds, which varies significantly and systematically with frequency, magnitude, contact area, and stimulation location [17]. Detection thresholds however alone are not enough to describe the tactile sense [14].

print and read: [18]

2.1.2 Comparison with Auditory Perception

Merchel et al. [19] compiled a comparative survey of psychophyical aspects of auditory and tactile perception. In real life, perception of such stimuli are often coupled, and a better understanding may allow design of may allow novel multi-modal systems that match perceptually. Though their anatomy and physiology differ considerably, the two senses share many similarities.

Intensity / dB

Sound is the sound pressure level, Lspl. It is defined as the logarithmic ratio of the effective value of the sound pressure p and has a reference value p0 = 20µ Pa: Lspl = 20 log p/p0 dB.

A similar unit for measuring vibrations is the acceleration level Lacc. It is defined as the logarithmic ratio of the acceleration a and has a reference value a0 = 1µm/s2: Lacc = 20 log a/a0 dB. In contrast to sound pressure level, 0dB acceleration level is not related to the perception threshold.

Sensitivity

Auditory: "It can be seen that they get closer toward lower frequencies. The audi-tory dynamic range is thus frequency dependent from 50dB to more than 100dB."

Tactile: "the glabrous skin becomes more sensitive to the acceleration of its surface with decreasing frequency" "sensitivity depends on the dis- tribution and density of the mechanoreceptors, with lower thresholds for areas with higher receptor density [56]. Hairy skin is approximately 10–20dB less sensitive depending on frequency [101]" "Again a frequency dependence can be seen, with smaller dynamic ranges for frequencies above approximately 300Hz" "to [63]where anewperceptually motivatedmeasurementwas proposed to represent human vibration intensity perception: the perceived vibration magnitude M in vip, comparable to auditory loudness N in sone."

Age

" At higher frequencies, where the Pacinian system is predominant, a strong loss of sensitivity can be observedwith increasing age. No effectwas found for low frequencies."

Energy integration

"An other important characteristic of the auditory and vibro- tactile modality, which has an influence on the threshold, is the ability to integrate energy. This is often discussed using the relationship between the duration and the threshold (or intensity) of a stimulus." "The auditory threshold of detection decreases with increasing duration up to a stimulus length of approximately 1s." "Tactile Temporal energy integration can also be found in the vibrotactile domain, but only in the Pacinian system [32,33]." "No temporal summation was found for low frequencies, e.g. below 25 Hz" "measured a 3dB reduction of threshold per doubling of duration up to a stimulus length of 300ms, indicating a complete integration of energy. Similar curves were found at 100Hz and 500Hz, frequencies at which mainly the Pacinian corpuscles are responsive to vibration. The same trend was found in suprathreshold experiments [3]. Other experiments by the author with seat vibrations at 40Hz, 80Hz, 160Hz and 320Hz confirmed the above con-clusions but are not plotted here for clarity" " Using smaller contact areas, more and more non-Pacinian receptors will be stimu-lated [86]. Consequently, the amount of temporal summation declines." " Evidence for energy integration within the Pacinian channel has been dis-cussed above and addition of sensation magnitudes between mechano-receptive channels has been reported [60,104]. It was therefore suggested that the Pacinian channel is analo-gous to a critical band in the auditory system [57]."

[rest not yet condensed/read]

2.1.3 Perceptual Dimensions

RT Verrillo seems to be the guy here. Brewster defined basic tacton dimensions, todo: complete (2004 at least).

[20] Haptical Phonemes: “building blocks”, combined with concatenation and superposition

Duration [21]

Location Very salient, but not easily varied. For belts, generally four or more actuators; for balance feedback, Sienko et. al(Add Citation: *ref*) showed that there was no benefit of 8 actuators over four.

“Pulses of different durations can encode infor- mation. Gunther (2001) investigated a range of subjective responses to pulses of different durations. He found that stimuli lasting less than 0.1 seconds were perceived as taps or jabs whereas stimuli of longer duration, when combined with gradual attacks and decays, may be per- ceived as smoothly flowing tactile phrases. He suggests combining duration with alterations in the envelope of a vibration, e.g. an abrupt attack feels like a tap against the skin, a gradual attack feels like something rising up out of the skin.” [22] “Duration and tempo Tactons are not mapping dependent Longer vibrations resulted in an increase in perceived magnitude of all data concepts. Similarly, a faster tempo of a repeating vibration resulted in an increase in perceived magnitude across all data concepts. ” [23]

Amplitude Between level of detection and painful (55dB). Lower threshold varies with age [24], inter-subject (best to let users choose themselves [25]!), and with menstruation cycle [26].

Roughness Prominently researched in the context of tactons: Amplitude modulated sinusoids (e.g., 250 Hz * 50 (rough) / 30 Hz (very rough)), [21]

Frequency “While humans can hear sounds in the range 20-20,000Hz, the practical frequency range of the skin is much smaller, ranging from 10Hz to 400Hz . The usable frequency range is further reduced by the limited bandwidth of the devices used (see Section 2.1), making frequency modulation alone unusable in Tacton design with these devices. ” . . . “Using amplitude as a parameter in Tactons could be equally problematic as reducing the amplitude could degrade perception of other parameters, or render the signal undetectable, while increasing it too far could cause pain [7]. Therefore, it is best to leave amplitude under the control of the user. ” [25] Depends on skin type [27] “A range of frequencies can be used to differentiate Tactons. The range of 20 – 1000 Hz is perceivable but maximum sensitivity occurs around 250 Hz (Gunther et al., 2002). The number of discrete values that can be differentiated is not well understood, but Gill (2003) sug- gests that a maximum of nine different levels can be used. As in audition, a change in amplitude leads to a change in the perception of frequency so this has an impact on the use of frequency as a cue. The number of levels of fre- quency that can be discriminated also depends on whether the cues are presented in a relative or absolute way. Mak- ing relative comparisons between stimuli is much easier than absolute identification, which will lead to much fewer discriminable values, as shown in the work on ear- con design (Brewster et al., 1994).” [22]

10-400 ([21]) or 1000(?) Hz. Buzzing, feeling of periodicity below 100 Hz, smooth above, “were: 6 Hz (slow motion, very rough), 70Hz (fluttering slightly faster motion, rough), and 250 Hz (smooth) as shown” [21]

Waveform “Initial pilot studies with 6 participants showed they could distinguish between a sine wave (very fast, smooth), and sawtooth wave (fast, rough), and a square wave (slow, very rough). Therefore, in this ex- periment a sine wave, square wave and sawtooth wave were used. The square waves were created using the Fourier series made up of the sum of odd harmonics of sine waves. When adding harmonics, it was ensured that the amplitude levels created by each harmonic were always within the 250Hz resonating frequency of the actuator…the design of Tactons. The results, with recognition rates of 94.2% for differing waveforms, 81% for differ- ing frequencies, and 61.1% for amplitude modulation indicate that users can identify and distinguish differing waveforms significantly more effectively than amplitude modulation and frequency. Therefore, different wave- forms can be used as the texture parameter in Tacton design. Previous Tacton design has used amplitude modulation to create the roughness parameter but accuracy was not high enough for reliable use. Given that using differing waveforms produces high recognition rates, by changing the technique used to create texture, overall recognition rates for 3 parameter Tactons could be increased” [21]

Morphing Proposed by UBC folks: [20], [28] implemented on motor-based rig

Rhythm “Rhythm is an extremely important parameter in earcon design and could be equally, if not more, important in Tactons. Rhythms can be created by grouping together pulses of different durations. Summers [8] used temporal patterns (rhythms) along with frequency and amplitude to encode speech information in vibrations, and found that participants mainly used the information obtained from the temporal patterns, rather than from the frequency/amplitude modulation. This suggests that rhythm could be an important parameter in Tacton design.“ [25]

2.1.4 Tactile Illusions

The fundamental limits of tactile information processing, however, are still largely unknown [29]. It is likely that there are a number of “haptic illusions” yet undiscovered. One such illusion relevant to the design of information displays is the “cutaneous rabbit”, or sensory saltation [30]. It describes the perception of a rapid sequential stimulus to be perceived as a gradual movement [31]. Several factors, such as the two-point discrimination threshold [2], the distance between actuators, inter-stimuli interval, saltatory area, and repetition, affect this sensory saltation [30], [31]. Borst and Baiyya demonstrated that people can interpret saltation sequences as strokes on a two-dimensional tactile display [32]. The Tactile Brush algorithm described by Israr et al. provides a generalized way to compute such strokes on an actuator grid [33].

2.2 Biofeedback and Augmented Feedback

Supporting the Perception-Action Loop

2.2.1 Definitions

The human brain has been shown to be extraordinary in its ability to be plastic and adapt to new challenges. Given appropriate stimulation, rehabilitation for even grave damage can sometimes achieved. This *neuroplasticity* is the basis for so called *sensory substitution*. Coupling information from an artificial receptor to the brain via a human-machine interface to replace a lost sense, we can learn to use this information transparently in place of that usually transmitted from an intact sense organ [34]. Sensory substitution can be either permanent, replacing the lost sense with an apparatus that is always used when the sense is needed, or temporary when such a device is used only during rehabilitation [35]. Examples of the permanent case are the tactile visual sensory substitution devices [36]–[38] for visual function and cochlea implants for auditory function [39]. It has been shown that the original senses are preserved and the additional sense does not lead to information overload, that is regular touch and the substitution stimuli are not confused [34], [40]. Substituting the vestibular system with accelerometer or plantar pressure readings relayed to the skin is an example for temporary sensory substitution [41]. Sensory substitution is thought to support the (re-)learning of appropriate input weights for the sensorimotor integration loop, in this case ignoring a damaged vestibular organ and prioritizing visual and haptic sensation.

If the existing sense is not to be replaced, but to be supported with additional information such external feedback, received in addition to the task-specific, intrinsic type during an activity, is called *sensory augmentation*, *augmented feedback* [42] or *biofeedback*. The terms will be used interchangeably in the following. Augmented feedback can either provide knowledge of result, that is how close the user got to achieving the goal, or knowledge of performance, that is about the quality of the movement itself [42].

Biofeedback means relaying measurements of a target biomedical variable to the user, either directly as a numeric reading or indirectly through transformed feedback [43]. Examples for the direct approach are fitness watches displaying the instantaneous heart rate, and the “quantified self”-style long-term metric tracking. The indirect approach, that is, transforming the readings into another modality, makes up the majority biofeedback research and our review in the following.

Biofeedback systems range from purely passive devices to systems that provide the user with entirely new senses [34], [44], and have been shown to be effective as support for a number of pathologies and can improve motor learning and performance [42]. Further, augmented feedback can help guide the execution specific movements, both for in rehabilitation and regular motor learning [45], [46]. The neurological mechanisms for this are still under investigation; Basmajian suggested that new pathways might be developed in the brain as a result of using biofeedback [47], [48]. The general principle of a biofeedback system is to map measurements of (e.g., postural) parameters on an output, thus providing corrective feedback and closing a perception-action loop. In the case of balance training, augmented feedback has been demonstrated to reduce postural sway [48]–[50], regardless of feedback modality (i.e., vibrotactile, electrotactile, auditory, visual).

On the passive end, it has been demonstrated that just wearing stimulating insoles with spikes, aimed at increasing the plantar cutaneous sensation, can significantly enhance balance [51]. Similarly, insoles with rims improve the perception of the limit of stability [52]. More active, insoles generating subsensory mechanical noise [53] can enhance plantar tactile sensation based on the principle of stochastic resonance.

2.2.2 Modalities and Applications

(ToDo: *briefly arch over range of applications, then give overview for applications in sensorimotor rehabiliation. This section is meant to be the general introduction, we only go into detail on balance later*) In rehabilitation research, the general goal of biofeedback is to provide cues in real time to help patients regain their sensorimotor abilities. These cues inform the user about an instantaneous error corresponding to body limb kinematics, kinetics, or neuromuscular activity in the context of a physical task [48].

pneumatic for gait in peripheral neuropathy [54] for prostheses [55]

Allison M. Okamura: soft robotics and pneumatic/fabric+servo feedback

Auditory

Sonification, the non-verbal auditory representation of data, allows perceiving typically non-audible information by using sound that is systematically bound to the underlying data [56](ToDo: *better sonification reference*) and has been successfully used used previously for sports training [57], [58]. The auditory perception is especially suitable as communication channel if the eyes are not available such in our scenario, where the visual perception is usually dominated by fixating a reference point to stabilize the posture.

2.3 Wearables

[59] The Design of Wearable Technology: Addressing the Human-Device Interface through Functional Apparel Design

[60] The Invisible future: The seamless integration of technology into everyday life

2.4 Vibrotactile Actuators

2.4.1 Evolution and Overview

Gault's work in the 1920s on auditory sensory substitution [61] represents some of the earliest research in haptic rendering, building on equipment loaned to him by the Bell telephone company to relay speech sound to vibrations on the hand. Technology for rendering haptic stimuli has advanced considerably since Gault's pioneering work in the 1920s [61]. Research continues to investigate the fundamental principles of haptic perception [11], [12], but generally lags behind the comprehensive understanding of the human visual and auditory perception.

Here, we focus on vibrotactile perception and describe a variety of present actuator, or *tactor*, technologies. Generally, we can distinguish between tactors that provide a low degree of control over frequency vs. amplitude, and those that provide a higher "fidelity" or definition.[2] Both have their place in the current stage of vibrotactile research wearable research.

2.4.2 Eccentric rotating mass

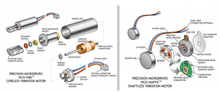


Figure 2.2: Eccentric Rotating Mass Motor (ERM) (ToDo: *own figure*)

The most common type of haptic actuator and ubiquitous in its use in mobile phones, eccentric rotating mass motors (ERMs) come in various sizes and strengths and are available either as a cylindrical or a pancake-shaped unit (E and P in Figure 2.1, respectively). ERMs are the most inexpensive but fundamentally constrained in their control, as rotation / vibration frequency is proportional to vibration amplitude. Furthermore, ERMs need around five revolutions to reach the intended amplitude(ToDo: *fig/citation*), and similarly need more time to stop. This means that shorter stimuli are not possible. ERMs normally are brushed DC motors, and as the brushes deterioate the life-span of ERMs is limited. (Add Citation: *reference on erm limited lifespan*)

[2] HD Haptics Specification: `https://github.com/HapticsIF/HDActuatorSpec`

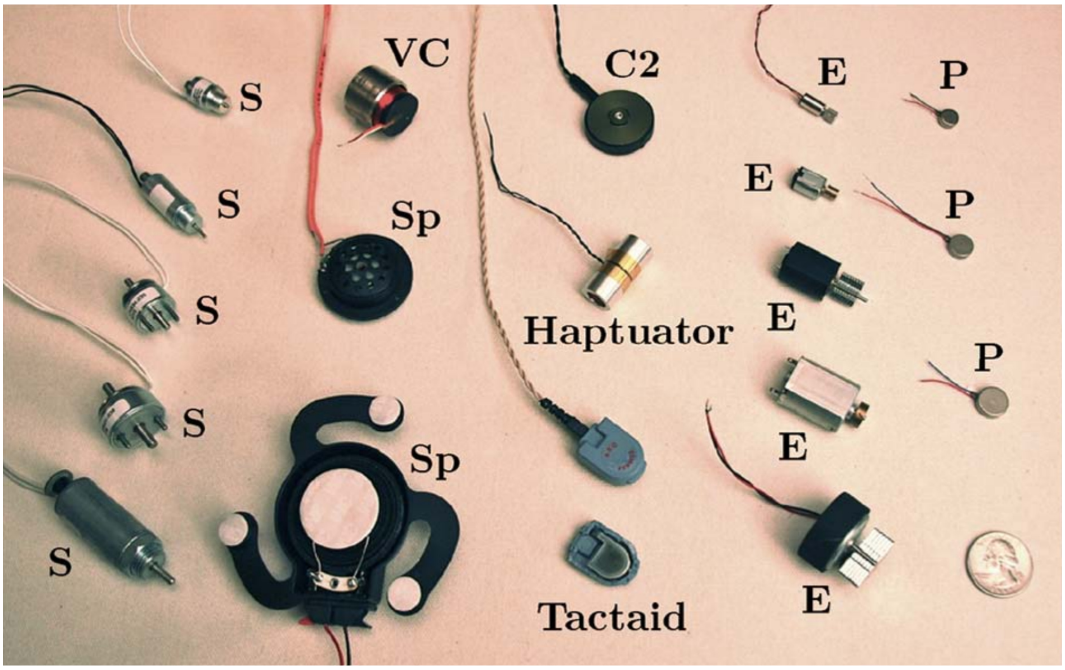


Figure 2.1: Sample vibrotactile actuators: S: Five solenoids of varying sizes. VC: A commercial voice coil without bearings. Sp: Two audio speakers. C2: A C2 tactor from EAI. Haptuator: A Haptuator mkI from Tactile Labs, Inc. Tactaid: One complete Tactaid from AEC and one opened to show the suspension inside. E: Five shafted/cylindrical eccentric rotating mass motors. P: Three shaftless/pancake eccentric rotating mass motors. A U.S.quarter appears at bottomright for scale. Figure and caption by Choi and Kuchenbecker [1].

2.4.3 Voice coils

Voice coils are a type of solenoid (S in Figure 2.1), which consist of a coil enclosing a movable piece of ferromagnetic material suspended between springs. In voice-coils (VC, SP, Haptuator, Tactaid, C2 in Figure 2.1), the movable core is a magnet [1]. As the coil can be driven at arbitrary frequencies, independent control of frequency and amplitude is possible. Voice coils connected to a suspended membrane form loudspeakers. The improved control and quicker response to transients of voice coil tactors consequently is similar to loudspeakers and are able to render a variety of effects that are impossible with ERMs. One example is the switch-like force-feedback to replace the physical switch on touchpads, as demonstrated with Apple's Taptic engine (Figure 10 in the patent [62, p32], `Illustratedathttps://www.ifixit.com/Teardown/Retina+Macbook+2015+Teardown/39841#s89855`).

Wide-band voice coil actuators can provide a more uniform frequency response and represent what we consider "high-fidelity" actuators, as they can be used to render much more complex stimuli. They operate on more than one fixed frequency, and can respond to quicker stimuli (below 0.5s) On the DIY end, it is possible to strip the voice coil from a loudspeaker to create such an actuator, power-to-weight ratio and integration are not very good though. Commercially, there are few options, mostly targeted at low volumes for researcher purposes. We discuss two examples here, the Haptuators sold by Tactile Labs, and Engineering Acoustics C2 and C3 tactors.

Stimulus Synthesis: SuperCollider, Max and PureData

Tactile Labs Haptuators

The Tactile Labs Haptuators [63] are a good example to illustrate the technological evolution of such actuators: Figure 2.3 shows the original Haptuator (TL002-14-A), it's redesign (TL002-1R), and the latest Haptuator mkII (TL002-09-A), with much higher actuation strength in a significant smaller package. These actuators are designed for research, retail around 250 USD for the Haptuator mkII at the time of writing and are only produced in limited quantities. (ToDo: *Clark Synthesis!*)

Together with ERMs, the C2/C3 tactors voice-coil actuators, are most common research projects to date. (ToDo: *needs a place?*)

C2 Tactors

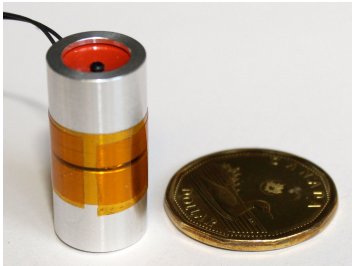
- common mass-produced tactor for rnd purposes
- delivers force through indentor pin to the body
- expensive, but very robustly made to be used stand-alone
- resonance peak around peak resonance frequency [64] around 190 Hz, Fig. 2.4fig:c2lra

Linear Resonant Actuators

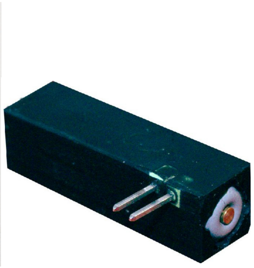
Linear resonant actuators (LRAs), shown in Figure 2.4, are a newer type of actuator based voice coils that is beginning to replace the small ERMs in consumer electronics, e.g., smartphones, due to a number of benefits. LRAs are operated at a peak resonant frequency (commonly around 200 Hz), hence the name, rendering them more power efficient than ERMs for delivering simple buzzes. Response time is, similar to other voice coils, much quicker than with ERMs. As there are not motors with brushes, durability is much improved over ERMs. Unlike ERMs, LRAs produce a linear motion, again meaning that more power can be oriented towards the user.[3]

The frequency response of LRAs exhibits a clear resonance frequency, similar to that of the C2 tactors, as shown in Fig. 2.4 2.4c, albeit with significantly weaker maximum acceleration [64], which is also reflective of the size differences. ERMs and LRAs are designed for the mass market and retail at less than $10 per actuator and represent the state of the art of what we consider “low-fidelity” tactors.

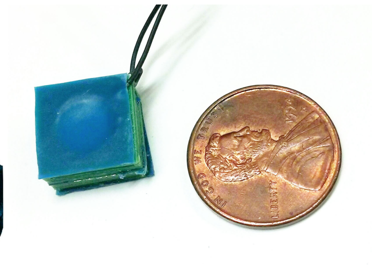
[3]Texas Instruments claims 50% more force at 50% less power than an ERM (`https://www.ti.com/lit/ml/sszb151/sszb151.pdf`).



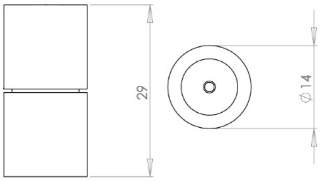
(a) Original Haptuator



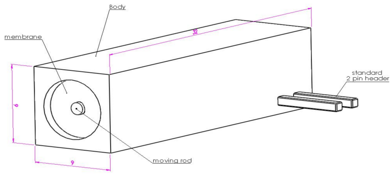
(b) Haptuator mkII



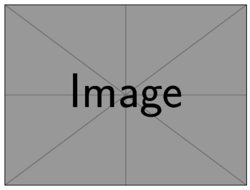
(c) Haptuator Planar



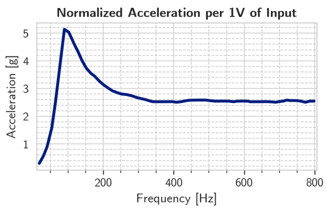
(d) Haptuator Original



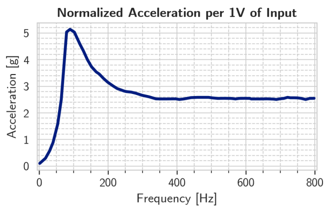
(e) Haptuator mkII



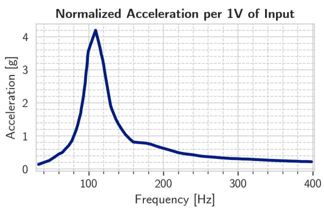
(f) Haptuator Planar



(g) Haptuator Original

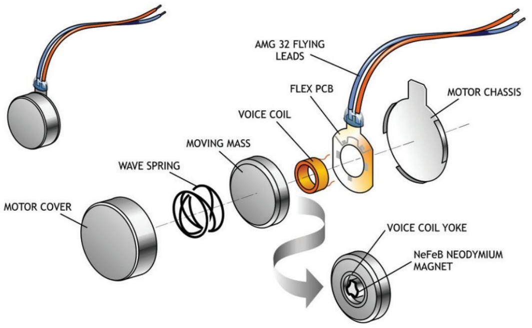


(h) Haptuator mkII

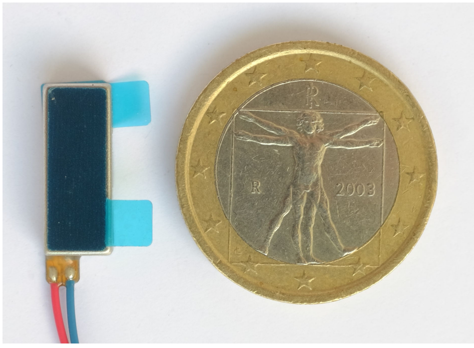


(i) Haptuator Planar

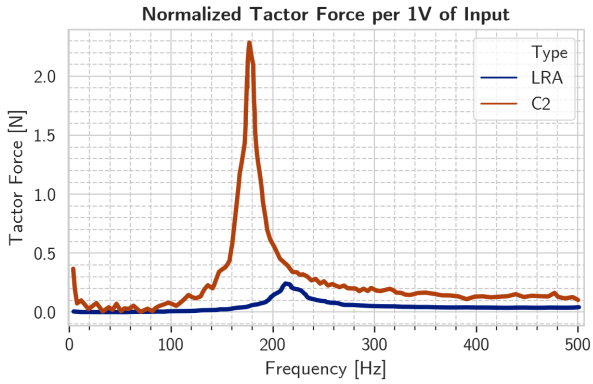
Figure 2.3: Three Tactile Labs Haptuators, their dimensions and frequency responses (notice the different scales). Figures taken from Tactile Labs website and datasheets thereon (http://www.tactilelabs.com).



(a) Coin-cell LRA. Figure taken from Precision Microdrives `http://precisionmicrodrives.com`



(b) Horizontal LRA, Jin Long LVM061530B



(c) Frequency Response of C2 Tactor vs. LRA (unspecified model); measured here were the forces against a static object. Figure adapted from Azadi et al. [64].

Figure 2.4: Linear Resonant Actuators

2.4.4 Tactor Drivers

ERMs are powered with DC voltage as regular brushed motors, solenoids and voice coils are driven with AC input signal. Control of ERMs thus can be implemented with simple means, e.g., a transistor on the output pin of a microcontroller, optionally controlling internal rotation speed by varying the duty cycle of a PWM output. The drive signal for LRAs is commonly a square wave with the resonance frequency of the actuator, e.g., 205 Hz.

The Texas Instruments DRV2605[4] driver chip we chose for the designs presented in this thesis can operate both ERMs and LRAs. It provides advanced features that improve and facilitate actuator control in comparison to the other options available on the market at design time. ERMs can be ramped up and braked faster through overshooting respectively short-circuiting the outputs and tracking the back-EMF of motors. The resonance tracking feature adjusts the drive frequency to the actual resonance frequency, which depends on the integration in to the housing and varies slightly per actuator unit and is measured on startup. This ensures optimal actuations strength. The DRV2605 is controlled through an I2C serial interface, and thus does not require a PWM-capable I/O pin on the microcontroller for each actuator that is to be controlled. It provides an internal library of 128 tactile effects, but can also receive intensity control input in various forms, e.g., over I2C. The chip can be powered directly from a battery voltage and provides uniform drive strength attuned to the parameters of the actuator. In terms of space required for driver electronics, the chip is similar to what would be required to drive a pin through a transistor.

For voice coils, the input signal is similar to an audio signal, and thus synthesis and drive amplification can be achieved with technology for audio applications. This means anything from a PWM-based synthesis approach, such as Mozzi (ToDo: *ref*), to a full desktop computer running SuperCollider or MaxMSP/Puredata for synthesis, and audio amplifiers ranging from headphone amplifiers to speaker amplifier stages, depending on power rating of the actuator.

[4] `http://www.ti.com/product/drv2605`

2.4.5 Other Vibrotactile Actuator Types

Piezoelectric

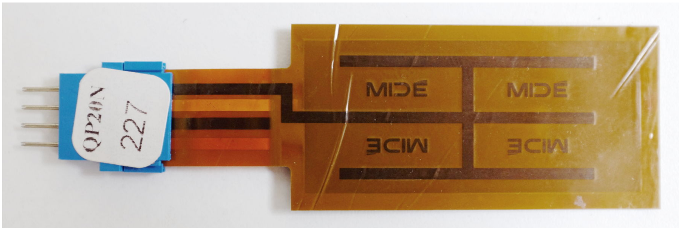


Figure 2.5: Midé QP20N, Figure taken from http://www.mide.com

Another tactile actuator technology makes use of the piezoelectric effect, which deforms certain crystals when a voltage is applied across them. This effect is reversible, and piezoelectric elements can also be used to measure deformation and perform energy harvesting. Piezo elements are lightweight and low profile. Although their deformation is small, they can be used to render very crisp effects. As high-bandwidth actuators, they allowing independent control of amplitude and frequency, and are driven with an AC voltage, typically in the range of 200 V. Stacked actuator types, such as the Midé[5] QP20N sensor/actuator, shown in Figure 2.5, provide the highest forces.

The per-unit price of the illustrated actuators ranges from less than ten dollars for ERMs and LRAs, $75 for the Midé QP20N, to $250 for an Haptuator mkII.

Electrovibration

The cutaneous touch organs can also be stimulated directly with electrical current, resulting in a sensation described as a tingle, itch, vibration, buzz, touch, pressure, pinch, and sharp and burning pain, depending on the stimulating voltage, current, and waveform [4]. The obvious advantage of electrotactile stimulation is that no mechanical displacement is involved, thus maximizing the achievable bandwidth. Some work on hand-held [65] and forehead-worn [66]

[5] http://www.mide.com/products/quickpack/quickpack-piezoelectric-actuators-and-sensors.php

devices exists, but establishing reliable electric contact to the afferent nerves or receptors remains a challenge. Skin resistance varies with thickness and hydration. Consequently, depending on the contact location, voltages as high as ±600V may be required [67]. This motivates placing the electrodes on the tongue, where the receptors are close to the surface and the mucous skin provides good electrical contact. As an example, the tongue display unit (TDU), an array of electrotactile electrodes in a tongue retainer [36], [40], [68], was used for rehabilitation applications, notably visual and vestibular sensory substitution.

Further, an effect known as electrovibration occurs when a dry patch of skin moves over an electrode surface excited with a low frequency AC voltage. This effect is believed to be due to electrostatically generated mechanical skin deformation [69], [70] and is being explored commercially to extend touch screen displays.

(ToDo: *RPP deduplication/missing check!*)

2.4.6

2.5 Sensing Postural Parameters

The basis for all proposed applications is reliable sensing of postural parameters. These could be single parameters, such as trunk orientation, or a set of parameters, such as the joint angles of a skeletal model. In the following sections we review the most relevant sensor technologies. We follow with an overview of clinical measures for assessing the posture and movement of patients in Section 3.1.2. These are the measures which we ultimately hope to improve with our proposed technology.

Generally, there are two broad categories for sensing motion: sensors worn directly on the body, and systems that track movement from a distance. Optical motion capture (MoCap) is an example for the latter.

2.5.1 Optical Motion Capture

We distinguish two general approaches to optical motion capture. The first works by placing a reflective marker on the body, illuminating the scene with infrared light, and tracking these highly contrasting reference points with high-speed cameras. Examples for this approach are the systems by Vicon[6] and Qualisys[7] on the high-end and the NaturalPoint Optitrack[8] on the low-end of the price-performance range. Marker-based systems require an often lengthy and sometimes challenging process of attaching markers onto the body or object that is to be measured. Nevertheless, these systems remain the gold standard in motion tracking, offering high temporal and spatial accuracy. Companion software packages can fit skeletal models that provide joint angles, center of body mass, and absolute movement trajectories that commonly serve as ground truth for validating other kinds of motion tracking **MartinaManciniLaurieKing2013**.

Markerless optical motion capture do not require markers and use intensity images, depth images or a combination of both. The probably most common example is the Microsoft Kinect, developed as a motion capture accessory for the XBox game console. The original Kinect version uses a depth sensor made by Primesense that provides the depth computation in the hardware. The Primesense sensor is based on a structured light approach: a fine, known pattern of laser light is cast onto the scene, and a camera records the deformation of this pattern. In contrast, the more recent Kinect Version 2 relies on a Time-of-Flight (ToF) depth sensor instead. Computing the depth information based on the return time of light emitted from a source in the unit, this version has reduced latency and increased resolution and field of view. Being aimed at the consumer market, Kinect sensors are inexpensive, and come with an SDK that provides a machine-learning based skeletal fitting. Since its introduction in 2010, Kinects have been used in a number rehabilitation applications [71].

Optical motion capture inherently relies on an unobstructed line-of-sight between the body/object at all times, and thus problems with occlusion,

[6] http://www.vicon.com
[7] http://www.qualisys.com
[8] http://www.naturalpoint.com/optitrack/

reflections and ambient lighting are common. Using only a single camera, the Kinect can only reasonably track users facing the camera; at oblique angles, the tracking breaks down due to self-occlusion of the body. Multiple cameras can to be used to overcome this and track all degrees of freedom, and this is how aforementioned marker-based systems work. The necessary extrinsic calibration and system cost associated with commercial MoCap systems normally prohibits their use outside of laboratory conditions. Another type of markerless motion capture system is based on body-worn sensors, introduced in the next section.

2.5.2 MARG (Magnetic, Angular Rate and Gravity) Sensors

Body-worn motion sensors have become much more affordable and compact in recent years [72] and may be sufficiently accurate for our application. Important characteristics of such sensors are bandwidth, noise floor, cross-axis sensitivity, drift, linearity, dynamic range, shock survivability, and power consumption[9].

Sensing Posture: Duplicate info from belt section

We plan to use a combination of accelerometers, gyroscopes and magnetometer sensors as the basis for sensing the relevant, task-dependant postural parameters of the user. These sensors are common in current consumer devices and their quality and availability continues to increase. For static balancing tasks, a three-axis accelerometer reading can be sufficient to infer the user's upper-body tilt, as for example demonstrated by Tyler et al. in their balance rehabilitation system for vestibular loss. [41]. When we asked slackline instructors about the postural parameter they give the most feedback on in beginner slacklining, they identified bending the knees to lower the center of gravity and recenter it over the base of support as the primary strategy of which they reminded beginners. One initial question we will address is whether we can compute the knee bend angle reliably from two accelerometer measurements, with the sensors worn on the thigh and calf, or if we have to use other types of sensors such as goniometers.

[9] http://www.sensorsmag.com/sensors/acceleration-vibration/an-overview-mems-inertial-sensing-technology-970

Goniometers can be made out of potentiometers and flexible sticks [73] and yield drift-free angle readings. Another challenge arises in integrating the readings from multiple sensors and fusing them into an orientation estimate to compensate for the accuracy issues of each individual sensor type, such as drift for gyroscopes. Commercially available motion tracking systems based on such sensors, from companies such as Xsens,[10] are accompanied by software packages that provide sensor fusion and fitted skeletal models. The result is a stream of motion tracking data similar to optical motion capture systems such as the ones sold by Vicon,[11] but designed to operate outside of lab environments and without stationary, calibrated cameras.

Sensor Types

We plan to use such a sensor-based system to understand which specific measurements are appropriate and necessary to track postural parameters during balance exercise, then implement these measures in a custom system. Further, utilizing a proven industrial system will allow us to focus mostly on the feedback part of our problem in the beginning. Two main sensor types often combined into so called inertial measurement units (IMUs) are *accelerometers*, measuring acceleration, and *gyroscopes*, measuring angular velocity. A three-axis accelerometer, when not otherwise accelerated, will always measure the earths gravity and this vector can be used as an absolute orientation value. During motion, relative orientation angles can be computed by integrating gyroscope measurements. Over time however, measurement error accumulates [74]. Combining accelerometers and gyroscopes, we can use the gravity vector as a reference to compensate for this *drift*, but only for two of the axis, the ones parallel and perpendicular to gravity. A third, non-inertial sensor type that is often combined with the first two to address this yaw drift are *magnetometers*, also referred to as compass sensors. Measuring the surrounding magnetic field, they can provide an absolute heading to the earth's north pole. However, the natural magnetic field is often distorted by ferromagnetic objects, rendering magnetometer readings on their own unreliable.

[10] `http://www.xsens.com/`
[11] `http://www.vicon.com`

This combination of *magnetic, angular rate and gravity (MARG sensors)* sensors, has become the most common type of sensor in consumer electronics, such as smartphones and tablets. Sensors for three-axis with digital readout are the norm, and on the consumer market end of MARG instrumentation are chips such as the Invensense MPU-9150 that combines three-axis sensing for all three types of sensors in a $4 \times 4 \times 1$ mm package. Sensor models with large ranges or high accuracy are not yet as miniaturized and only available for one or two axis measurements. Applications requiring this higher performance thus commonly still rely on multiple discrete sensors [75].

Manufactured as microelectromechanical (MEMS) systems, MARG sensor price and size has decreased significantly in the last years.[12] MEMS accelerometers consist of a mass-spring system in a vacuum, and measure the linear acceleration by measuring the displacement of this mass, often through capacitive sensing. MEMS gyroscope often use capacitive sensing as well, but measure the displacement of a vibrating mass suspended in two orthogonal sets of springs, thus exploiting the coriolis effect.[13] MEMS magnetometers commonly measure the magnetic field, for example through the Lorentz force acting on a current-carrying conductor or the voltage invoked by the hall-effect.

MARG-based Motion Tracking Systems

Each sensor type has intrinsic shortcomings and combining the output of multiple sensors is necessary to obtain robust and accurate orientation readings. Many algorithms for this sensor fusion have been proposed and implemented[74], such as Kalman [48], [76] and other filters [77], as well as machine learning techniques [78]. Commercial MARG-based MoCap systems, such as the ones sold by Xsens,[14] list their integrated Kalman filtering as one of the strong selling points. Xsens products include both wired and wireless systems, ranging from single sensor units to whole-body suits capable of providing skeletal segment

[12]The Invensense MPU-9150 retails for less than $10 per single unit.
[13]http://www.sensorwiki.org/doku.php/sensors/gyroscope
[14]http://www.xsens.com

tracking similar to the optical MoCap systems, but without requiring external instrumentation. The price is similarly high.[15]

Another notable MARG-based motion sensing product is the APDM Mobility Lab[16]. This system is aimed at supporting physio- and occupational therapist in assessing mobility in an ambulatory context. The system consists of wearable, wireless synchronized sensor nodes. The primary aim of this system is to instrument standard assessment tests, such as the Timed Up and Go test, and it has been validated for this application. https://learn.adafruit.com/adafruit-bno055-absolute-orientation-sensor/device-calibration

Consumer Electronics Systems Combining MARG and Optical Tracking

There are two notable tracking systems used in rehabilitation research based on consumer gaming hardware that combine optical tracking with motion sensing. Released in 2006, the controller of the Nintendo Wii console contains a three-axis accelerometer (and three-axis gyroscope in the Plus model released in 2010). An on-chip tracking of infrared lights from a "sensor bar" that is placed on top of the game display, allows the controller to be used as pointing device. The Sony Playstation 3 Eye camera[17] tracks the illuminated orb of the Playstation Move controller,[18] which in turn contains three-axis accelerometer, gyroscope and magnetometer sensors.

2.5.3 Pressure Based Sensing

A third class of instrumentation measures the pressure of users onto the ground. Maintaining balance means deriving appropriate responses to the gravito-inertial environment, and consequently, measuring the force exerted by a subject on the ground can reveal a lot of information on the balance performance and gait

[15] As an example, the full body MVN Biomech Awinda system for human kinematic measurements retails at USD 74,000.00 (Price as of January 2014, consisting of 18 MTw sensors, receiver station, straps and a reference video camera).

[16] http://apdm.com/gait-and-posture/Mobility-Lab/

[17] http://us.playstation.com/ps3/accessories/playstation-eye-camera-ps3.html

[18] http://us.playstation.com/ps3/accessories/playstation-move-motion-controller-ps3.html

characteristics. There are three broad classes of sensing devices: instrumented shoes/insoles, force plates, and pressure mapping systems.

Instrumented Shoes and Insoles

Instrumented shoes and insoles most commonly employ a number pressure sensors, such as force sensing resistors (FSRs), placed underneath the points of highest pressure contact. These include the heel (medially and laterally) and the ball behind the toes (under both the first and fifth metatarsal heads) [54], [79]–[88]. Instead of discrete pressure sensors, Kong and Tomizuka introduced a shoe that measures air pressure [89]. The pressure sensors are often in combined with bend and inertial motion sensors.

At least two commercial products provide sensing based on above general four-pressure sensor idea. The discontinued Nike Hyperdunk+[19], aimed at sharing movement statistics with peers, combines four pressure sensors and an accelerometer. Similar is a a recent Kickstarter [20]. The Moticon OpenGo Science[21] was developed for clinical research and sports science and combines 13 pressure sensors that can communicate through the ANT wireless protocol with a host computer [90]. Finally, going even further in the number of integrated sensors are grids of sensors. Wu and Chen presented a grid of 87 piezoresistive sensors for instrumenting footwear [90]. Companies such as Tekscan[22] and Novel [23] offer insole-shaped versions of their pressure mapping systems that provide the highest available spatial resolution. They are discussed in their own section.

Force Platforms

Force platforms, or *force plates*, are platforms resting on one or more force sensors, such as strain gauges, piezoelectric or piezoresistive sensors. If they are in a single-pedestal configuration, that is instrumented with only a single sensor,

[19] http://www.nike.com/ca/en_ca/c/basketball/nike-basketball-hyperdunk-plus
[20] https://www.kickstarter.com/projects/universole/universole-pressure-tracking-insoles
[21] http://www.moticon.de/en/products/opengo-science
[22] http://www.tekscan.com/medical/system-fscan1.html
[23] http://www.novel.de/novelcontent/pedar

they can only measure the vertical component of the force on the platform. Multi-pedestal configurations, with three or four sensors, are capable of sensing the movement of pressures across the surface of the platform and are commonly used to measure movements, such as walking. Being normally laboratory-grade equipment, force platforms usually have a price of several thousand dollars.

There is one comparable consumer electronics device. Introduced in 2007 as an accessory to the Wii Console, the Wii Balance board[24] has been shown to be an affordable, portable, valid and reliable alternative for balance assessment [91], [92] and has since found many applications in rehabilitation, especially for serious games [93].

Pressure Mapping Systems

Increasing the number of pressure sensors even further are pressure mapping systems, consisting of a grid of interconnected FSR, piezoresistive or capacitive sensors. Offered by companies such as Tekscan[25], they provide a high sensing element density of almost four sensing elements per square centimeter. Tekscan offers both mats for placement on the floor and insole-shaped versions for placement in shoes. Aimed at clinicians, these systems are priced similarly to force platforms. A recently announced example of a pressure mapping system for in-home use is the instrumented yoga mat SmartMat[26]. It is aimed at guiding users through posture based on the pressure distribution of their hands and feet with a smart phone application.

Joint Angles

Goniometry, the measurement of angles, is most commonly achieved with one of two approaches: with a potentiometer or flexible strain gauges attached between the limbs, or by comparing the readouts of two MARG sensors [94]. The former can be bulky and is prone to alignment problems, which may make the use of MARG sensors more practical. MARG sensing is typically

[24] http://www.wiifit.com/what-is-wii-fit-plus/#balance-board
[25] http://www.tekscan.com/medical.html
[26] http://smartmat.com/

accurate to within a few degrees, both for research systems and commercial specifications, for instance from XSens and APDM [95].

(ToDo: *Bend Sensors*)

2.5.4 Other Sensors

Electromyography

Electromyography (EMG) is based on recording and evaluating the electrical activity by the muscles upon their activation, which provides information on the actual muscle activations. Both surface and intramuscular systems exist, the latter requiring subdermal electrodes that, depending on the legislation, may only be placed by physicians. Surface EMG requires the skin to be shaven and cleaned, often also abrased to improve contact and adhesion of the electrodes, and is limited in reliability due to the distance to deeper muscles. The raw electrical measurements have to be amplified and the superimposed motor unit action potentials from several muscles have to be decomposed. Biofeedback on EMG readings was shown to be effective in the treatment of many musculoskeletal conditions and in post cardiovascular accident rehabilitation [43]. Recently, Thalmic Labs introduced the Myo[27], a mass-market input gesture recognition device using EMG readings to detect five hand gestures.

EEG

Time-of-Flight

2.6 Embedded Computing

Introduce different tiers of computing, e.g., ATMega vs Raspberry Pi

[27]https://www.thalmic.com/en/myo/

2.6.1 Microcontroller Architectures

Atmel ATMega 16/32/128/256: 8-Bit

initially: ATMega 168/328/32u4 (16/32 KiB of Flash, 16 Mhz), Arduino Mega / Wiring Board w/ ATMega128 (expensive, big)

Arm Cortex M-Profile: 32-Bit

- Exists for a long time, but not easily accessible to amateur developers: high pin count chips, very heterogenic; Arm licenses IP cores and individual manufacturers produce the actual chips including memories, clocks, and interfaces for communication (e.g., ADCs, DACs, USB, Bluetooth).
- Very heterogeneous: Arm microcontrollers range from super-energy efficient cores to fully fledged CPUs for personal computers (form the basis for basically all smart phones today); Linux kernel has support for xxx architectures alone. Here: two examples that form the basis for systems developed in this thesis.
- Here: Cortex Arm v6 / v7 m architectures, 32 Bit
- Come in different "profiles", A for high performance, R for real-time, M for mobile, energy efficient (here). Optional extensions (Armv7: floating point, DSP)
- Armv7 is optimized for deeply embedded applications and as
- Armv6 is subset of Armv7, predecessor
- basis for most consumer electronics, from printers to smart phones
- generally more capable microcontrollers, actively developed and forms the basis of many Atmel chips (the XMega series never became common in the Arduino world, expensive, high density chips and specialized, for many fast PWM channels, for example)
- Recent years have seen the adoption of ARM MCUs in the Arduino world; Arduino Due for example, but also many boards by third parties

ARM Cortex M0

- M0: low-power, minimal silicon surface, can be roughly compared to ATMega328 (16 Mhz), but much more flash (256/521 KiB)
- Example: Nordic nRF51: Ultra-Low power, demos run on a coin cell, integrates BLE stack
- Example: Arduino M0, Adafruit Feather M0
- Example: Inside Bosch BNO055: used for sensor fusion

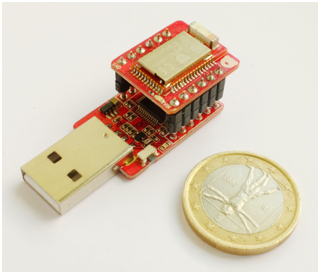


Figure 2.6: BLE Nano with Programmer

Arm Cortex M4F

- M4F: newer ARMv7-M architecture, larger instruction sets, DSP instructions, floating point unit (F), 64 Mhz, more interrupts, optimized for low cycle count operations
- uses more power than M0, but can be more power efficient overall: FPU, SIMD, sleeps more: " The ARM Cortex-M4 with its SIMD and floating-point capabilities ran the tests 12 to 174 times (FPU!) faster than the ARM Cortex-M0 core and consumed 2x to 9x more power. Consequently, the ARM Cortex-M4 core proved to be more energy efficient than the ARM Cortex-M0 core."[28]
- Example: Teensy Kinesis K20

[28] http://www.ganssle.com/tem/tem228.html Retrieved 01.07.2019

- Example: Nordic nRF52832: M4F core successor: much faster clock (48 Mhz?), Bluetooth 5 support, Ant

2.6.2 Prototyping Systems

Arduino

- Massively abstracted from the pains of low-level programming
- Atmel MCU, as opposed to, e.g., Microchip PIC, used GCC as compiler, making it immediately accessible for open source and cross-platform development tools
- Processing-based minimalistic IDE and sketch-approach, together it became the go-to solution for beginner embedded programming (before: Basic Stamp for hobbyists)
- Wiring language
- Created an ecosystem of hardware manufacturers and library developers (common business model: we give you the library, please buy our prototyping boards)

(ToDo: *Maybe add comparison table between ATMega and the two Arms*)

Teensy 3+

The Teensy 3.2[29] is based on a NXP Kinesis MK20DX256VLH7 MCU, running at 72 Mhz with 256 KiB flash memory and 32 KiB RAM. It features a large number of capable I/O ports: 16-bit ADC, a 12-bit DAC, and among others support for the IC-IC-sound (I2S) protocol.(ToDo: *move to technologies*) This means, it has hardware support to connect to standard audio DACs/codecs. The Kinesis K20 sub-family also comprises DSP instructions that enable signal processing in hardware.

[29] https://pjrc.com/teensy/techspecs.html

Stackable

https://tinycircuits.com/pages/tinyduino-overview

$BRIX_2$

The presented systems are representative of most commercial motion sensing in that they only support sensing, mostly through propriate software packages, as opposed to providing a platform that could be extended, both for additional sensors as well as to deliver localized feedback. A number of systems presented by the DIY and research communities over the years package MARG sensors for prototyping and wearable applications, but few are extensible.

This shortcoming motivated the development of the $BRIX_2$ [96],[30] and we will discuss the $BRIX_2$ system as an example.

The $BRIX_2$ system and some extensions are shown in Figure 2.7.

The $BRIX_2$ stand out as a modular system specifically designed for developing wireless sensor/actuator networks in ubiquitous computing research. The base module of the current version incorporates two Atmel ATMega 8-bit microcontrollers and an Invensense MPU-9150 motion sensor chip (three-axis accelerometers, gyroscopes, and magnetometers). The MPU-9150 can perform on-chip sensor-fusion of its accelerometer and gyroscope readings to provide an absolute orientation measurement. An onboard low-power wireless interface enables communication among several $BRIX_2$ units to form a body-worn sensor network. The system is Arduino-compatible, facilitating rapid development and giving access to a wide range of existing software libraries.

Normally housed in a 4×8 LEGO® brick, each base module has three slots for extension modules, in 2×4 bricks each, that stack onto it. Many extensions already exist, such as an eccentric-mass vibrator motor module and a Bluetooth Low Energy (BTLE) module to connect the body worn components to a smartphone.

[30] http://www.techfak.uni-bielefeld.de/ags/ami/brix2/

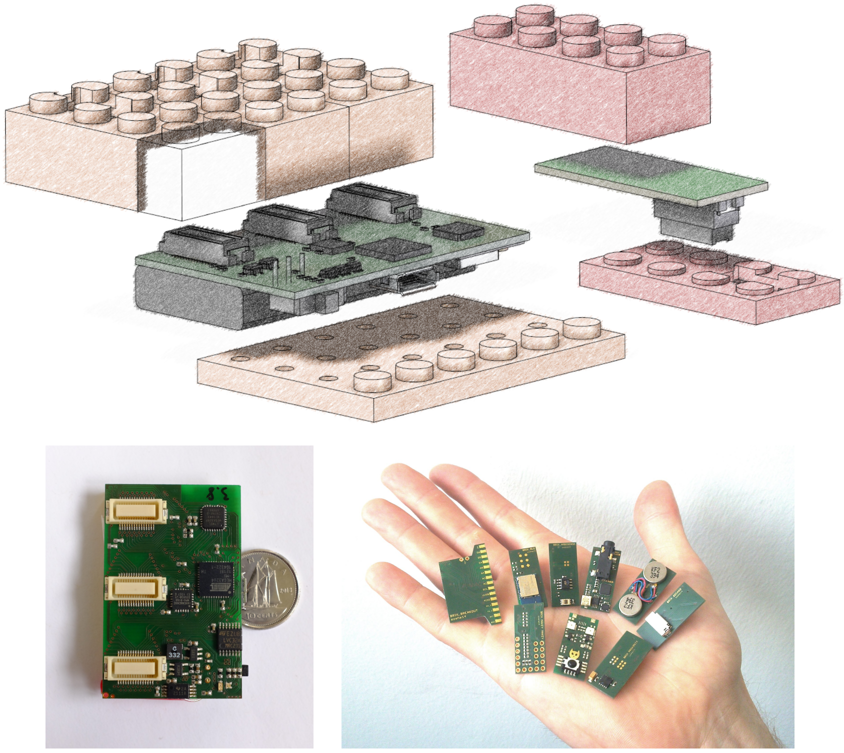


Figure 2.7: The BRIX$_2$ system: Explosion drawing, base module without casing next to a quarter coin for size reference, a selection of extension modules, and a chest strap mount. Figures taken from BRIX$_2$ website, `http://tiny.cc/brix2`.

Raspberry Pi

I2C-based Prototyping Systems

Sparkfun Qwiic, Elecrow Crowtail, Seeedstudio Grove

Xu and Sienko: wearable sensing and feedback system

[97] proof-of-concept design of a configurable, wearable sensing and feedback system for real-time postural balance and gait training targeted for home-based treatments and other portable usage. Sensing and vibrotactile feedback are performed via eight distributed, wireless nodes or “Dots” (size: 22.5 × 20.5 × 15.0 mm, weight: 12.0 g) that can each be configured for sensing and/or feedback according to movement training requirements. In the first experiment, four healthy older adults were trained to reduce medial-lateral (M/L) trunk tilt while performing balance exercises. When trunk tilt deviated too far from vertical (estimated via a sensing Dot on the lower spine), vibrotactile feedback (via feedback Dots placed on the left and right sides of the lower torso) cued participants to move away from the vibration and back toward the vertical no feedback zone to correct their posture. A second experiment wasconducted with thesamewearablesystemtotrain sixhealthy older adults to alter their foot progression angle in real-time by internally or externally rotating their feet while walking

to be extended: Vibropixels etc

2.6.3 Communication

We need to communicate between wearable systems, either with other devices on the body, or with external, often stationary systems. For example, we may want to record sensor data for later analysis.

Bluetooth Low Energy (BLE)

In the case of devices on the body, Bluetooth Low Energy (BLE) has emerged as the de-facto standard for such devices and is natively supported by all current smart phones. BLE devices can either take a 'central' or a 'peripheral' role; depending on the device, both may be possible. Smartphones are usually a central role. In recent years, Web browsers started to include BLE control capabilities through the WebBluetooth API, which means that werable controller applications can be implemented as websites, that work across platforms, e.g., on an Apple iOS and Android phones/tablets and on a desktop computer. As an evolution of Bluetooth classic it is currently at version 5 of the protocol, which means it is featureful but also quite complex. Its main feature is security (albeit hard to implement correctly due to the complexity), and compatibility. Its downsides are high latency and jitter and comparably little bandwidth.

Network Communication

If we can use network communication, usually over wireless LAN, we can achieve much higher throughput and lower latency. Two protocols were used in this thesis: Open Sound Control (OSC) for low-latency realtime applications and LabStreamingLayer (LSL) for accurately time-synchronized data.

Open Sound Control (OSC) Open Sound Control (OSC) is a simple network protocol originally intended for communication among computers, sound synthesizers, and other multimedia devices [98].[31] It is based on UDP, which, as opposed to TCP, does not guarantee the delivery of packets at the benefit of being very low-latency. OSC wraps basic data types (32 Bit integer/float, strings, blobs) with a high-resolution timestamp and a path specifying the type of transmission. A number of implementations exist for most programming language, making it easy to implement as a user. Even implementations for embedded systems such as the ESP32 exist. It is very suitable for communicating control inputs, as in closed loop feedback systems, less so for recording data.

LabStreamingLayer (LSL) Lab Streaming Layer[32] is a more recent development originating in the biomedical domain, and is commonly used, for

[31] http://opensoundcontrol.org/
[32] https://github.com/sccn/labstreaminglayer

example, to transmit data from Electroencephalograpy (EEG) sensors in Brain-Computer-Interfaces (BCI). It is based on TCP, and uses multicast to announce available ”streams“ to receivers. More than one machine can receive data from a single stream source, timestamps between systems can be synchronized to achieve sub-millisecond accuracy on a local network of computers through a protocol similar to the Network Time Protocol (NTP).

LSL stream data contains a metadata field, that for example details channel labels and data units; each stream can contain a number of channels of one data type. Streams are commonly saved as XDF files that contain all metadata and timestamps. A cross-platform library for C++ exists and wrappers for most popular programming languages exist, making implementation straightforward in most cases. As a high-level protocol relying on the boost library, it is beyond what embedded systems can support.

2.7 Overview of VTW Applications

2.7.1 Notifications

2.7.2 Tactons

!!! lots of lit on this

“structured messages that can be used to communicate to users non-visually” [22]
icon, earcon, tactons [21]

“specific and short abstract tactile messsages that can easily be interpreted by users with minimal cognitive effort.” [28]

Awareness

Military/Aeronautics?

Navigation

2.7.3 Medical

Sensory Substitution / Augmentation

Gait Cueing

2.7.4 Entertainment

Rumble Pack - from Motors to LRAs

2.7.5 Control

Military/Aeronautics?

Alarm Systems

2.7.6 Design tools

2.8 Analysis: State of the Art and Our Approach to Improve on it

2.8.1 Shortcomings of the literature

crappy one-off prototypes: not reproducible, not robust
  low fidelity - comparable to just having one tone available
  not modular, or not adaptable to various applications
  not open source
  expensive or not modifieable

2.8.2 What applications do we want to support?

Movement Rehabiliation / Physio-/Occupational Therapy
  Augmented Feedback

Teaching

Student Projects

HCI Explorations

Multi-Tactor Notifications

Gestural Control

Long-Term Studies

2.8.3 Proposed Approach

Do focussed case studies

Balance Rehabilitation as a closed-loop feedback example

Involves VT at Hip

Sensing at Feet

Explore VT on Feet

References for Chapter 2

[1] S. Choi and K. J. Kuchenbecker, "Vibrotactile Display: Perception, Technology, and Applications," *Proceedings of the IEEE*, vol. 101, no. 9, pp. 2093–2104, Sep. 2013, ISSN: 0018-9219. DOI: 10.1109/JPROC.2012.2221071.

[2] M. F. Nolan, "Limits of two-point discrimination ability in the lower limb in young adult men and women.," *Physical therapy*, vol. 63, no. 6, pp. 492–499, Sep. 1983, ISSN: 0031-9023. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/6611663%20http://ptjournal.apta.org/content/63/9/1424.short.

[3] C. Gu and M. J. Griffin, "Vibrotactile thresholds at the sole of the foot: effect of vibration frequency and contact location.," *Somatosensory & Motor Research*, vol. 28, no. 3-4, pp. 86–93, Jan. 2011, ISSN: 1369-1651. DOI: 10.3109/08990220.2011.622493. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/22115026.

[4] K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita, and W. J. Tompkins, "Electrotactile and Vibrotactile Displays for Sensory Substitution Systems," *IEEE Transactions on Biomedical Engineering*, vol. 38, no. 1, pp. 1–16, 1991, ISSN: 15582531. DOI: 10.1109/10.68204. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/2026426.

[5] M. A. Nurse and B. M. Nigg, "Quantifying a relationship between tactile and vibration sensitivity of the human foot with plantar pressure distributions during gait," *Clinical Biomechanics*, vol. 14, no. 9, pp. 667–672, 1999, ISSN: 0268-0033. DOI: http://dx.doi.org/10.1016/S0268-0033(99)00020-0. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0268003399000200.

[6] Y. Visell, "Walking on virtual ground: physics, perception, and interface design," PhD Thesis, McGill University, 2011. [Online]. Available: http://gradworks.umi.com/NR/77/NR77568.html.

[7] M. Alfuth and D. Rosenbaum, "Effects of changes in plantar sensory feedback on human gait characteristics: a systematic review," *Footwear Science*, vol. 4, no. 1, pp. 1–22, Mar. 2012, ISSN: 1942-4280. DOI: 10.1080/19424280.2011.653990. [Online]. Available: http://www.tandfonline.com/doi/abs/10.1080/19424280.2011.653990.

[8] A. a. Priplata, J. B. Niemi, J. D. Harry, L. a. Lipsitz, and J. J. Collins, "Vibrating insoles and balance control in elderly people.," *Lancet*, vol. 362, no. 9390, pp. 1123–4, Oct. 2003, ISSN: 1474-547X. DOI: 10.1016/S0140-6736(03)14470-4. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/14550702.

[9] H. G. Kang, A. M. Galica, A. A. Priplata, O. V. Starobinets, S. E. D'Andrea, J. J. Collins, L. A. Lipsitz, F. A. Sorond, and L. A. Cupples, "Sub-threshold noise to the feet reduce gait variability in the elderly," *Journal of the American Geriatrics Society*, vol. 56, no. 4, S184–S184, 2008. DOI: 10.1016/j.gaitpost.2009.07.005.Subsensory.

[10] L. a. Jones and N. B. Sarter, "Tactile Displays: Guidance for Their Design and Application," *Human Factors: The Journal of the Human Factors and Ergonomics Society*, vol. 50, no. 1, pp. 90–111, Feb. 2008, ISSN: 00187208. DOI: 10.1518/001872008X250638. [Online]. Available: http://hfs.sagepub.com/cgi/doi/10.1518/001872008X250638.

[11] V. Hayward, "Is There a 'Plenhaptic' Function?" *Philosophical Transactions of the Royal Society B*, vol. 366, pp. 3115–3122, Nov. 2011.

[12] Y. Visell, B. L. Giordano, G. Millet, and J. R. Cooperstock, "Vibration Influences Haptic Perception of Surface Compliance During Walking," *PLoS ONE*, vol. 6, no. 3, e17697, 2011. DOI: 10.1371/journal.pone.0017697. [Online]. Available: http://dx.doi.org/10.1371/journal.pone.0017697.

[13] S. J. Bensmaïa and M. Hollins, "Complex tactile waveform discrimination," *The Journal of the Acoustical Society of America*, vol. 108, no. 3, p. 1236, 2000, ISSN: 00014966. DOI: 10.1121/1.1288937.

[14] S. R. Mills, M. Morioka, and M. J. Griffin, “Limitations of Vibrotactile Thresholds,” no. September, 2018.

[15] M. Hollins and S. R. Risner, “Evidence for the duplex theory of tactile texture perception,” *Perception and Psychophysics*, vol. 62, no. 4, pp. 695–705, 2000, ISSN: 00315117. DOI: 10.3758/BF03206916.

[16] Z. Halata and K. Baumann, “Anatomy of receptors,” in *Human Haptic Perception: Basics and Applications*, M. Grunwald, Ed., Basel: Birkhäuser, 2008, ch. 6, pp. 85–92.

[17] N. G. Forta, M. J. Griffin, and M. Morioka, “Vibrotactile difference thresholds: Effects of vibration frequency, vibration magnitude, contact area, and body location,” *Somatosensory and Motor Research*, vol. 29, no. 1, pp. 28–37, 2012, ISSN: 08990220. DOI: 10.3109/08990220.2012.662182.

[18] K. S. Hale and K. M. Stanney, “Deriving Haptic Design Guidelines from Human Physiological, Psychophysical, and Neurological Foundations,” *IEEE Computer Graphics and Applications*, vol. 24, no. 2, pp. 33–39, 2004, ISSN: 02721716. DOI: 10.1109/MCG.2004.1274059.

[19] S. Merchel and M. E. Altinsoy, “Psychophysical comparison of the auditory and tactile perception: a survey,” *Journal on Multimodal User Interfaces*, vol. 14, no. 3, pp. 271–283, 2020, ISSN: 17838738. DOI: 10.1007/s12193-020-00333-z. [Online]. Available: https://doi.org/10.1007/s12193-020-00333-z.

[20] M. Enriquez, K. E. MacLean, and C. Chita, “Haptic phonemes: Basic building blocks of haptic communication,” *ICMI’06: 8th International Conference on Multimodal Interfaces, Conference Proceeding*, pp. 302–309, 2006. DOI: 10.1145/1180995.1181053.

[21] E. Hoggan and S. A. Brewster, “New parameters for tacton design,” *Conference on Human Factors in Computing Systems - Proceedings*, pp. 2417–2422, 2007. DOI: 10.1145/1240866.1241017.

[22] S. A. Brewster and L. M. Brown, "Tactons: structured tactile messages for non-visual information display," Australian Computer Society, 2004. [Online]. Available: http://crpit.com/Vol28.html.

[23] J. Ferguson, J. Williamson, and S. A. Brewster, "Evaluating mapping designs for conveying data through tactons," *ACM International Conference Proceeding Series*, pp. 215–223, 2018. DOI: 10.1145/3240167.3240175.

[24] R. T. Verrillo, S. J. Bolanowski, and G. A. Gescheider, "Effect of aging on the subjective magnitude of vibration," *Somatosensory and Motor Research*, vol. 19, no. 3, pp. 238–244, 2002, ISSN: 08990220. DOI: 10.1080/0899022021000009161.

[25] L. M. Brown, S. A. Brewster, and H. C. Purchase, "A first investigation into the effectiveness of Tactons," *Proceedings - 1st Joint Eurohaptics Conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems; World Haptics Conference, WHC 2005*, pp. 167–176, 2005. DOI: 10.1109/WHC.2005.6.

[26] G. A. Gescheider, R. T. Verrillo, J. T. Mccann, and E. M. Aldrich, "Effects of the menstrual cycle on vibrotactile sensitivity," *Perception & Psychophysics*, vol. 36, no. 6, pp. 586–592, 1984, ISSN: 00315117. DOI: 10.3758/BF03207520.

[27] B. J. P. Mortimer, G. A. Zets, and R. W. Cholewiak, "Vibrotactile transduction and transducers," *The Journal of the Acoustical Society of America*, vol. 121, no. 5, pp. 2970–2977, 2007, ISSN: 0001-4966. DOI: 10.1121/1.2715669. [Online]. Available: http://scitation.aip.org/content/asa/journal/jasa/121/5/10.1121/1.2715669.

[28] K. E. Maclean and M. Enriquez, "Perceptual Design of Haptic Icons," *Proceedings of EuroHaptics 2003*, no. July, pp. 351–363, 2003, ISSN: 10699384. DOI: 10.3758/BF03194057.

[29] A. Gallace, H. Z. Tan, and C. Spence, "The Body Surface as a Communication System: The State of the Art after 50 Years," *Presence: Teleoperators and Virtual Environments*, vol. 16, no. 6, pp. 655–676, Dec. 2007, ISSN:

1054-7460. DOI: 10.1162/pres.16.6.655. [Online]. Available: http://www.mitpressjournals.org/doi/abs/10.1162/pres.16.6.655.

[30] F. A. Geldard, *Sensory saltation: metastability in the perceptual world*, English. Hillsdale, N.J.; New York: Lawrence Erlbaum Associates ; Distributed by the Halsted Press Division of Wiley, 1975, ISBN: 0470295716 9780470295717.

[31] S. " Lee and T. Starner, "BuzzWear: Alert Perception in Wearable Tactile Displays on the Wrist," in *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems (CHI '10)*, ser. CHI, New York, NY, USA: ACM, 2010, pp. 433–442, ISBN: 978-1-60558-929-9. DOI: 10.1145/1753326.1753392. [Online]. Available: http://doi.acm.org.proxy2.library.mcgill.ca/10.1145/1753326.1753392.

[32] C. W. Borst and V. B. Baiyya, "A 2D haptic glyph method for tactile arrays : Design and evaluation," in *EuroHaptics conference, 2009 and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems. World Haptics 2009. Third Joint*, Mar. 2009, pp. 599–604. DOI: 10.1109/WHC.2009.4810862.

[33] A. Israr and I. Poupyrev, "Tactile Brush: Drawing on Skin with a Tactile Grid Display," in *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems*, ser. CHI '11, New York, NY, USA: ACM, 2011, pp. 2019–2028, ISBN: 978-1-4503-0228-9. DOI: 10.1145/1978942.1979235. [Online]. Available: http://dl.acm.org/citation.cfm?id=1979235%20http://doi.acm.org/10.1145/1978942.1979235.

[34] P. Bach-y-Rita and S. W. Kercel, "Sensory substitution and the human–machine interface," *Trends in Cognitive Sciences*, vol. 7, no. 12, pp. 541–546, Dec. 2003, ISSN: 13646613. DOI: 10.1016/j.tics.2003.10.013. [Online]. Available: http://linkinghub.elsevier.com/retrieve/pii/S1364661303002900.

[35] Y. Visell, "Tactile sensory substitution: Models for enaction in HCI," *Interacting with Computers*, vol. 21, no. 1-2, pp. 38–53, Jan. 2009, ISSN: 09535438. DOI: 10.1016/j.intcom.2008.08.004. [Online]. Available:

http://dl.acm.org/citation.cfm?id=1507761.1507842%20http://dx.doi.org/10.1016/j.intcom.2008.08.004.

[36] P. Bach-y-Rita, K. A. Kaczmarek, M. E. Tyler, and J. Garcia-Lara, "Form perception with a 49-point electrotactile stimulus array on the tongue: a technical note.," *Journal Of Rehabilitation Research And Development*, vol. 35, no. 4, pp. 427–430, 1998. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/10220221.

[37] P. Bach-y-Rita, M. E. Tyler, and K. A. Kaczmarek, "Seeing with the Brain," *International Journal of Human-Computer Interaction*, vol. 15, no. 2, pp. 285–295, Apr. 2003, ISSN: 1044-7318. DOI: 10.1207/S15327590IJHC1502_6. [Online]. Available: http://dx.doi.org/10.1207/S15327590IJHC1502_6.

[38] P. Bach-y-Rita, C. C. Collins, F. A. Saunders, B. White, and L. Scadden, "Vision Substitution by Tactile Image Projection," *Nature*, vol. 221, no. 5184, pp. 963–964, Mar. 1969. [Online]. Available: http://www.nature.com/nature/journal/v221/n5184/abs/221963a0.html%20http://dx.doi.org/10.1038/221963a0.

[39] B. C. Papsin and K. A. Gordon, "Cochlear implants for children with severe-to-profound hearing loss," *New England Journal of Medicine*, vol. 357, no. 23, pp. 2380–2387, 2007.

[40] K. A. Kaczmarek, "The tongue display unit (TDU) for electrotactile spatiotemporal pattern presentation," *Scientia Iranica*, vol. 18, no. 6, pp. 1476–1485, Dec. 2011, ISSN: 10263098. DOI: 10.1016/j.scient.2011.08.020. [Online]. Available: http://linkinghub.elsevier.com/retrieve/pii/S1026309811001702.

[41] M. E. Tyler, Y. Danilov, and P. Bach-Y-Rita, "Closing an open-loop control system: vestibular substitution through the tongue," *Journal of integrative neuroscience*, vol. 2, no. 2, pp. 159–164, 2003. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/15011268.

[42] B. Lauber and M. Keller, “Improving motor performance: Selected aspects of augmented feedback in exercise and health,” *European Journal of Sport Science*, Sep. 2012, ISSN: 1746-1391. DOI: 10.1080/17461391.2012.725104. [Online]. Available: http://www.tandfonline.com/doi/abs/10.1080/17461391.2012.725104.

[43] O. M. Giggins, U. M. Persson, and B. Caulfield, “Biofeedback in rehabilitation,” *Journal of neuroengineering and rehabilitation*, vol. 10, no. 1, p. 60, Jan. 2013, ISSN: 1743-0003. DOI: 10.1186/1743-0003-10-60. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3687555&tool=pmcentrez&rendertype=abstract.

[44] S. K. Nagel, C. Carl, T. Kringe, R. Märtin, and P. König, “Beyond sensory substitution–learning the sixth sense.,” *Journal of Neural Engineering*, vol. 2, no. 4, R13–26, Dec. 2005, ISSN: 1741-2560. DOI: 10.1088/1741-2560/2/4/R02. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16317228.

[45] J. Lieberman and C. Breazeal, “Development of a Wearable Vibrotactile Feedback Suit for Accelerated Human Motor Learning,” *Robotics and Automation, 2007 IEEE International Conference on*, no. April, pp. 4001–4006, 2007. DOI: 10.1109/ROBOT.2007.364093. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=4209711.

[46] J. van der Linden, E. Schoonderwaldt, J. Bird, J. V. D. Linden, W. Hall, and M. Keynes, “Good vibrations: Guiding body movements with vibrotactile feedback,” in *Proceedings of the Third International Workshop on Physicality*, Sep. 2009, pp. 13–18. [Online]. Available: http://oro.open.ac.uk/19251/.

[47] J. V. Basmajian, *Biofeedback : principles and practice for clinicians*, English. Baltimore: Williams & Wilkins, 1983, ISBN: 0683003569 9780683003567.

[48] B.-C. Lee, “Design and Assessment of Vibrotactile Biofeedback and Instructional Systems for Balance Rehabilitation Applications,” Ph.D. Thesis. University of Michigan, 2012.

[49] A. Zijlstra, M. Mancini, L. Chiari, and W. Zijlstra, "Biofeedback for training balance and mobility tasks in older populations: a systematic review.," *Journal of neuroengineering and rehabilitation*, vol. 7, no. 1, p. 58, Jan. 2010, ISSN: 1743-0003. DOI: 10.1186/1743-0003-7-58. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3019192&tool=pmcentrez&rendertype=abstract.

[50] S. Kok-soon, L. Ming-Yih, L. Chih-feng, and T. Wen-Wei, "A new trunk sway assessment protocol using biofeedback inertial-based sensing modality for stroke patients," in *System Science and Engineering (ICSSE), 2011 International Conference on*, Jun. 2011, pp. 675–678, ISBN: 9781612844718. DOI: 10.1109/ICSSE.2011.5961988.

[51] E. Palluel, I. Olivier, and V. Nougier, "The lasting effects of spike insoles on postural control in the elderly," *Behavioral neuroscience*, vol. 123, no. 5, pp. 1141–7, Oct. 2009, ISSN: 1939-0084. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/19824780.

[52] S. D. Perry, A. Radtke, W. E. McIlroy, G. R. Fernie, and B. E. Maki, "Efficacy and effectiveness of a balance-enhancing insole.," *The journals of gerontology. Series A, Biological sciences and medical sciences*, vol. 63, no. 6, pp. 595–602, Jun. 2008, ISSN: 1079-5006. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/18559634.

[53] A. A. Priplata, B. L. Patritti, J. B. Niemi, R. Hughes, D. C. Gravelle, L. A. Lipsitz, A. Veves, J. Stein, P. Bonato, and J. J. Collins, "Noise-enhanced balance control in patients with diabetes and patients with stroke.," *Annals of Neurology*, vol. 59, no. 1, pp. 4–12, Jan. 2006, ISSN: 0364-5134. DOI: 10.1002/ana.20670. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16287079.

[54] Z. McKinney, K. Heberer, B. N. Nowroozi, M. Greenberg, E. Fowler, and W. Grundfest, "Pilot evaluation of wearable tactile biofeedback system for gait rehabilitation in peripheral neuropathy," in *Haptics Symposium (HAPTICS), 2014 IEEE*, Feb. 2014, pp. 135–140, ISBN: 9781479931316. DOI: 10.1109/HAPTICS.2014.6775445. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6775445.

[55] R. E. Fan, M. O. Culjat, C.-H. King, M. L. Franco, M. Sedrak, J. W. Bisley, E. P. Dutson, and W. S. Grundfest, "A prototype haptic feedback system for lower-limb prostheses and sensory neuropathy.," *Studies in health technology and informatics*, vol. 132, no. 3, pp. 115–9, 2008, ISSN: 0926-9630. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/18391269.

[56] J. Anlauff, T. Hermann, T. Grosshauser, and J. R. Cooperstock, "Modular tacTiles for Sonic Interactions with Smart Environments," in *Proceedings of the 4. International Workshop on Haptic and Auditory Design (HAID 09)*, E. Altinoy, U. Jekosch, and S. Brewster, Eds., ser. Lecture Notes on Computer Science, vol. 2, Dresden, Germany: Springer, 2009, pp. 26–27.

[57] J. Hummel, T. Hermann, C. Frauenberger, and T. Stockman, "Interactive sonification of german wheel sports movement," in *Human Interaction with Auditory Displays–Proceedings of the Interactive Sonification Workshop*, 2010, pp. 17–22. [Online]. Available: http://interactive-sonification.org/ISon2010/proceedings/papers/HummelHermannFrauenbergerStockman_ISon2010.pdf.

[58] T. Hermann and S. Zehe, *Sonified Aerobics - Interactive Sonification of coordinated body movements*, 2011. [Online]. Available: http://pub.uni-bielefeld.de/publication/2277186.

[59] L. E. Dunne, "The Design of Wearable Technology: Addressing the Human-Device Interface through Functional Apparel Design," Master's Thesis, Cornell University, 2004, pp. 1–190.

[60] P. J. Denning, *The Invisible future: The seamless integration of technology into everyday life*. McGraw-Hill, 2002, ISBN: 0071382240. [Online]. Available: http://www.loc.gov/catdir/description/mh024/2001031508.html.

[61] R. H. Gault, ""Hearing" by touch; demonstration of a case," *The Laryngoscope*, vol. 37, no. 3, pp. 184–189, Mar. 1927, ISSN: 1531-4995. DOI: 10.1288/00005537-192703000-00004. [Online]. Available: http://dx.doi.org/10.1288/00005537-192703000-00004.

[62] J. T. Bernstein, A. Cieplinski, B. W. Degner, D. Kerr, P. Kessler, P. Puskarich, M. H. Coelho, and A. Pance, *Touch Pad with Force Sensors and Actuator Feedback*, 2013.

[63] H.-Y. Yao and V. Hayward, "Design and analysis of a recoil-type vibrotactile transducer.," *The Journal of the Acoustical Society of America*, vol. 128, no. 2, pp. 619–27, Aug. 2010, ISSN: 1520-8524. DOI: 10.1121/1.3458852.

[64] M. Azadi and L. a. Jones, "Vibrotactile actuators: Effect of load and body site on performance," *2014 IEEE Haptics Symposium (HAPTICS)*, pp. 351–356, Feb. 2014. DOI: 10.1109/HAPTICS.2014.6775480.

[65] M. E. Altinsoy and S. Merchel, "Electrotactile Feedback for Handheld Devices with Touch Screen and Simulation of Roughness," *IEEE Transactions on Haptics*, vol. 5, no. 1, pp. 6–13, Jan. 2012, ISSN: 1939-1412. DOI: 10.1109/TOH.2011.56. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=6042861.

[66] H. Kajimoto, Y. Kanno, and S. Tachi, "A vision substitution system using forehead electrical stimulation," in *ACM SIGGRAPH 2006 Sketches on - SIGGRAPH '06*, New York, New York, USA: ACM Press, Jul. 2006, p. 39, ISBN: 1595933646. DOI: 10.1145/1179849.1179898. [Online]. Available: http://dl.acm.org/citation.cfm?id=1179849.1179898.

[67] M. A. Schaning and K. A. Kaczmarek, "A high-voltage bipolar transconductance amplifier for electrotactile stimulation.," *IEEE transactions on bio-medical engineering*, vol. 55, no. 10, pp. 2433–43, Oct. 2008, ISSN: 1558-2531. DOI: 10.1109/TBME.2008.926675. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=2586184&tool=pmcentrez&rendertype=abstract.

[68] Y. Danilov and M. E. Tyler, "Research Report Brainport: An Alternative Input to the Brain," *Journal of Integrative Neuroscience*, vol. 4, no. 4, pp. 537–550, 2005.

[69] K. a. Kaczmarek, K. Nammi, A. K. Agarwal, M. E. Tyler, S. J. Haase, and D. J. Beebe, "Polarity effect in electrovibration for tactile display," *IEEE Transactions on Biomedical Engineering*, vol. 53, no. 10, pp. 2047–2054, 2006, ISSN: 00189294. DOI: 10.1109/TBME.2006.881804.

[70] O. Bau and I. Poupyrev, "REVEL: tactile feedback technology for augmented reality," *ACM Transactions on Graphics*, vol. 31, no. 4, pp. 1–11, Jul. 2012, ISSN: 07300301. [Online]. Available: http://dl.acm.org/citation.cfm?id=2185520.2185585.

[71] B. Lange, C.-Y. Chang, E. Suma, B. Newman, A. S. Rizzo, and M. Bolas, "Development and evaluation of low cost game-based balance rehabilitation tool using the Microsoft Kinect sensor.," in *Annual International Conference of the IEEE Engineering in Medicine and Biology Society.*, vol. 2011, Jan. 2011, pp. 1831–4. DOI: 10.1109/IEMBS.2011.6090521. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6090521.

[72] C. I. Wall, N. D. Lyford, K. H. Sienko, and M. D. Balkwill, "The design and development of a production prototype balance belt," *Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE*, vol. 2011, pp. 3524–8, Jan. 2011, ISSN: 1557-170X. DOI: 10.1109/IEMBS.2011.6090585. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/22255100.

[73] T. Grosshauser and T. Hermann, "Wearable Setup for Gesture and Motion based Closed Loop Audio-Haptic Interaction," in *Proceedings of the 16th International Conference on Auditory Display (ICAD 2010)*, International Community for Auditory Display, 2010, pp. 31–38. [Online]. Available: http://pub.uni-bielefeld.de/publication/2017222.

[74] P. B. Shull, W. Jirattigalachote, and X. Zhu, "An Overview of Wearable Sensing and Wearable Feedback for Gait Retraining," English, in *Intelligent Robotics and Applications SE - 44*, ser. Lecture Notes in Computer Science, J. Lee, M. Lee, H. Liu, and J.-H. Ryu, Eds., vol. 8102, Springer Berlin Heidelberg, 2013, pp. 434–443, ISBN: 978-3-642-40851-

9. DOI: 10.1007/978-3-642-40852-6_44. [Online]. Available: http://dx.doi.org/10.1007/978-3-642-40852-6_44.

[75] M. Lapinski, M. Feldmeier, and J. A. Paradiso, "Wearable wireless sensing for sports and ubiquitous interactivity," *2011 IEEE SENSORS Proceedings*, pp. 1425–1428, Oct. 2011. DOI: 10.1109/ICSENS.2011.6126902. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=6126902.

[76] B.-C. Lee, S. Chen, and K. H. Sienko, "A wearable device for real-time motion error detection and vibrotactile instructional cuing.," *IEEE transactions on neural systems and rehabilitation engineering : a publication of the IEEE Engineering in Medicine and Biology Society*, vol. 19, no. 4, pp. 374–81, Aug. 2011, ISSN: 1558-0210. DOI: 10.1109/TNSRE.2011.2140331. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/21511568.

[77] S. O. H. Madgwick, "An efficient orientation filter for inertial and inertial / magnetic sensor arrays," Tech. Rep., 2010. [Online]. Available: http://www.x-io.co.uk/open-source-ahrs-with-x-imu/.

[78] S. O. H. Madgwick, A. J. L. Harrison, and A. Vaidyanathan, "Estimation of IMU and MARG orientation using a gradient descent algorithm.," in *IEEE International Conference on Rehabilitation Robotics*, vol. 2011, Jan. 2011, p. 5 975 346, ISBN: 9781424498628. DOI: 10.1109/ICORR.2011.5975346.

[79] J. A. Paradiso, K. Hsiao, A. Y. Benbasat, and Z. Teegarden, "Design and implementation of expressive footwear," *IBM Systems Journal*, vol. 39, no. 3.4, pp. 511–529, 2000, ISSN: 0018-8670. DOI: 10.1147/sj.393.0511. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5387021.

[80] S. J. Morris and J. A. Paradiso, "Shoe-integrated sensor system for wireless gait analysis and real-time feedback," in *Engineering in Medicine and Biology, 2002. 24th Annual Conference and the Annual Fall Meeting of the Biomedical Engineering Society EMBS/BMES Conference, 2002. Proceedings of the Second Joint*, vol. 3, Oct. 2002, 2468–2469 vol.3,

ISBN: 0780376129. DOI: 10.1109/IEMBS.2002.1053379. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=1053379.

[81] A. Y. Benbasat, "A wireless modular sensor architecture and its application in on-shoe gait analysis," in *IEEE Sensors Conference 2003*, 2003. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=1279111.

[82] J. A. Paradiso, S. J. Morris, A. Y. Benbasat, and E. Asmussen, "Interactive therapy with instrumented footwear," *Extended abstracts of the 2004 conference on Human factors and computing systems - CHI '04*, p. 1341, 2004. DOI: 10.1145/985921.986059. [Online]. Available: http://portal.acm.org/citation.cfm?doid=985921.986059.

[83] S. J. M. Bamberg, A. Y. Benbasat, D. M. Scarborough, D. E. Krebs, and J. A. Paradiso, "Gait analysis using a shoe-integrated wireless sensor system.," *IEEE transactions on information technology in biomedicine : a publication of the IEEE Engineering in Medicine and Biology Society*, vol. 12, no. 4, pp. 413–23, Jul. 2008, ISSN: 1558-0032. DOI: 10.1109/TITB.2007.899493. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/18632321.

[84] P. Paulick, H. Djalilian, and M. Bachman, "StabilitySole: embedded sensor insole for balance and gait monitoring," *Digital Human Modeling*, pp. 171–177, 2011. [Online]. Available: http://link.springer.com/chapter/10.1007/978-3-642-21799-9_19.

[85] G. Chelius, C. Braillon, M. Pasquier, N. Horvais, R. P. Gibollet, B. Espiau, and C. Azevedo Coste, "A Wearable Sensor Network for Gait Analysis: A Six-Day Experiment of Running Through the Desert," *IEEE/ASME Transactions on Mechatronics*, vol. 16, no. 5, pp. 878–883, Oct. 2011, ISSN: 1083-4435. DOI: 10.1109/TMECH.2011.2161324. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5975220.

[86] P. Lopez-Meyer, G. D. Fulk, and E. S. Sazonov, "Automatic Detection of Temporal Gait Parameters in Poststroke Individuals," *Information Technology in Biomedicine, IEEE Transactions on*, vol. 15, no. 4, pp. 594–601, Jul. 2011, ISSN: 1089-7771. DOI: 10.1109/TITB.2011.2112773.

[87] C. B. Redd and S. J. M. Bamberg, "A Wireless Sensory Feedback Device for Real-Time Gait Feedback and Training," *IEEE ASME Transactions on Mechatronics*, vol. 17, no. 3, pp. 425–433, 2012. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6175961.

[88] M. J.-D. Otis and B.-A. J. Menelas, "Toward an augmented shoe for preventing falls related to physical conditions of the soil," *2012 IEEE International Conference on Systems, Man, and Cybernetics (SMC)*, pp. 3281–3285, Oct. 2012. DOI: 10.1109/ICSMC.2012.6378297. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=6378297.

[89] K. Kong and M. Tomizuka, "A gait monitoring system based on air pressure sensors embedded in a shoe," *Mechatronics, IEEE/ASME Transactions on*, vol. 14, no. 3, pp. 358–370, 2009. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=4813239.

[90] C. Wu and C. Chen, "The development and verification of a piezoresistive foot pressure recording device," in *Orange Technologies (ICOT), 2013 International Conference on*, 2013, pp. 95–98, ISBN: 9781467359368. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6521166.

[91] R. A. Clark, R. McGough, and K. Paterson, "Reliability of an inexpensive and portable dynamic weight bearing asymmetry assessment system incorporating dual Nintendo Wii Balance Boards," *Gait & Posture*, vol. 34, no. 2, pp. 288–291, 2011, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2011.04.010. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636211001378.

[92] W. Young, S. Ferguson, S. Brault, and C. Craig, "Assessing and training standing balance in older adults: A novel approach using the 'Nintendo

Wii' Balance Board," *Gait & Posture*, vol. 33, no. 2, pp. 303–305, 2011, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2010.10.089. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636210003814.

[93] B. Lange, S. Flynn, R. Proffitt, C.-Y. Chang, and A. S. Rizzo, "Development of an interactive game-based rehabilitation tool for dynamic balance training.," *Topics in stroke rehabilitation*, vol. 17, no. 5, pp. 345–52, 2010, ISSN: 1074-9357. DOI: 10.1310/tsr1705-345. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/21131259.

[94] Freescale, *Implementation of an Electrogoniometer*, 2014. [Online]. Available: http://www.freescale.com/webapp/sps/site/overview.jsp?code=784_LPBB_ELECGNIOMTR (visited on 10/15/2014).

[95] P. B. Shull, W. Jirattigalachote, M. A. Hunt, M. R. Cutkosky, and S. L. Delp, "Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention," *Gait & Posture*, vol. 40, no. 1, pp. 11–19, 2014, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2014.03.189. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636214002872.

[96] S. Zehe, T. Grosshauser, and T. Hermann, "BRIX - An Easy-to-Use Modular Sensor and Actuator Prototyping Toolkit," in *Proceedings of the 4th International Workshop on Sensor Networks and Ambient Intelligence 2012*, Lugano, Switzerland, 2012.

[97] J. Xu, T. Bao, U. H. Lee, C. Kinnaird, W. Carender, Y. Huang, K. H. Sienko, and P. B. Shull, "Configurable, wearable sensing and vibrotactile feedback system for real-time postural balance and gait training: Proof-of-concept," *Journal of NeuroEngineering and Rehabilitation*, vol. 14, no. 1, pp. 1–10, 2017, ISSN: 17430003. DOI: 10.1186/s12984-017-0313-3.

[98] M. Wright and A. Freed, "Open sound control: A new protocol for communicating with sound synthesizers," in *Proceedings of the 1997*

International Computer Music Conference, International Computer Music Association San Francisco, 1997, pp. 101–104.

3 Vibrotactile Balance Feedback Belt (I 75 %, S 10%)

(
words)

[1], [2] give us all the reasons for biofeedback, and for building tech to evaluate long-term, in-situ. needs to be cost-effective for researchers to do, likely lots of individualized applications. (ToDo: *next two sections need to be restructured into the chapter opener, and the opener of the subsections*)

Better balance and mobility help to prevent falls and thus potentially grave injury, especially among groups with chronic deficits, such as post-stroke patients and the elderly [3]. In physical and occupational therapy, balance deficits are usually addressed with a set of repetitive exercises adapted to the patient's everyday activities. Motor learning and rehabilitation can be improved by providing external or "augmented" feedback [4], which augments the task-specific, intrinsic feedback during an activity. Such augmented feedback systems have been widely investigated, mostly in a clinical context [5], [6]. For balance (re-)learning specifically, augmented feedback may be used to support the brain's natural plasticity in reweighing sensory inputs.

A driving rehabilitation motivation for most patients is to return their home environment as soon as possible [7], but non-adherence to the prescribed excercise regimen remains a large problem [8]. Further, upon returning home after in-clinic rehabilitation, many patients find themselves unable to continue performing their activities at the same level at which they were previously capable in the clinic [9].

Here, we investigate the design of a belt-based wearable posture feedback system designed to support balance learning with augmented feedback in a home environment, where the usually required stationary equipment is not available. Our goal is a self-contained exercise support system that provides more detailed feedback than existing feedback belts. Improved in-home rehabiliation may allow patients to return home earlier, while also providing them with guidance on the execution of their exercises. (ToDo: *this touches on the text in the intro chapter; remove redundancy and pick where we do this right*)

We consider such feedback belts as a primary case study for the design of our toolkit, allowing us to gather insight on construction of haptic wearables and effective feedback systems, providing the basis for later improving excercise adherence through gamification and patient-therapist communication.

(ToDo: *Population: where does this go?*) Stroke affects balance in complex ways; patients are often elderly and thus already exhibit deficits and the size and location of lesions vary widely. Ultimatively, the target population are motivated elderly and high-functioning post-stroke patients with chronic balance deficits.

Similar systems have been applied to other pathologies, e.g., patients with Parkinson's disease and peripheral neuropathy. The users should have progressed well enough in in-clinic therapy to continue their therapy at home with a regimen of physiotherapy exercises for continued sensorimotor rehabilitation addressing their specific deficits.

We evaluated our system with healthy adults in a balance task that is difficult for them, as the motor learning processes are believed to be similar between the two cases (ToDo: *citation needed*). For example, balance difficulty increases on compliant surfaces, e.g., thick foam [10].

3.1 Background

3.1.1 Balance

Howe et al. define balance as "a somewhat ambiguous term used to describe the ability to maintain or move within a weight-bearing posture without falling", breaking it further down into the three aspects steadiness, symmetry, and dynamic stability.

Steadiness describes the ability to maintain a given posture with minimal extraneous movement, or sway. Symmetry refers to equal weight distribution between body parts that are weight-bearing, for example the feet while standing. The term dynamic stability is used to describe the ability to move within a given posture without loss of balance [11], for example, standing up from a sitting position.

In order to maintain standing balance, the *center of (body) mass* (COM) must be kept or returned over its base of support, and this maintenance of equilibrium is known as "postural control". Even during quiet stance, the COM sways constantly within the base, and activities such as walking, sitting down and reaching require control of the COM within a moving base of support [12].

3.1.2 Balance Performance Measures

(ToDo: *include: Standard techniques from 2002 [13]*)

Methods to assess the balance ability of an individual, for example to demonstrate effects of an intervention, usually either measure a direct physical physical variable, such as sway, or the functional performance during a set of tasks. This can be done either with *functional* and *instrumented* tests, reviewed briefly here.

Measures have to be specific to the information that is required, and Hayes separates the measures relevant to stroke rehabilitation into these three (verbatim) [12], [14]:

1. Global, clinical measures of balance dysfunction, used primarily to guide therapy and to evaluate therapeutic outcomes.

2. Sophisticated instrumented perturbation techniques, used (a) to assess differential diagnoses (e.g., is it a vestibular or proprioceptive deficit?) or (b) to evaluate more specific therapeutic intervention.

3. Simple quantitative tests of static postural instability, used where intermediate levels of information suffice.

A common functional test is the 'Get Up and Go' test in which the subjects are instructed to stand up from sitting comfortably on a chair, walk 3 meters, turn and return to sitting, while a therapist rates their performance on a five-point scale and optionally times the subjects. This simple test has been found to be test-retest reliable and to correlate with more sophisticated measures. Another common example are timed walking tests of 5, 10 or 30 meters.

Qualitative Measures

The first category of clinical assessment is most commonly done through protocols that instruct the person to be assessed to perform a set of movements, while the therapist watches, measuring time and rating on observation scales. These tests have the advantage to be time-efficient(ToDo: *this may not really be true in 2019*), low in cost and do not require special equipment or lengthy training [12].

Examples:

Timed balance tests These tests require individuals to balance under challenging conditions, such on one leg. In the simplest and oldest, the Romberg Test (RT) developed in 1851, a subject has to stand with their feet together, their eyes open and then closed and for 60 seconds each, while the examiner records the perceived amount sway under both conditions [12]. The RT is a very subjective and unreliable measure, and only tests the ability to stand with eyes closed. A modified, Sharpened Romberg Test, requires the patients to stand in *tandem stance* (heel-to-toe). The Single Limb Stance Timed Test (SLSTT) is another variation, in which the individuals have to stand on one leg for 30 seconds, with eyes either open or closed. It was shown that these tests have a ceiling effect for healthy adults.

1. Global, clinical measures of balance dysfunction, used primarily to guide therapy and to evaluate therapeutic outcomes.

2. Sophisticated instrumented perturbation techniques, used (a) to assess differential diagnoses (e.g., is it a vestibular or proprioceptive deficit?) or (b) to evaluate more specific therapeutic intervention.

3. Simple quantitative tests of static postural instability, used where intermediate levels of information suffice.

The Sensory Organization Test (SOT) evaluates somatosensory, visual, and vestibular function for upright equilibrium. In the SOT, duration of normal stance for normal vision and floor conditions, eye closed, vision manipulated for visual conflict, and somatosensory conflict (foam surface) conditions is recorded [12]. All these tests have shortcomings in the information they can collect, might not be suitable for some patients, and thus serve as simple screening tests that are best combined with other tests [12].

Rating Scales The Fugl-Meyer Assessment examines physical functioning after stroke, including a seven-item balance sub-section [15]. These cover sitting and standing balance, and the test has been shown to have a high inter-rater reliability and validity [12]. Widely used to test a static and dynamic balance abilities is the Berg Balance Scale (BBS), considered to be the gold-standard for functional balance tests (ToDo: *needs citation, [16] maybe*). It takes 15–20 minutes and consists of a set of 14 simple balance related tasks [17], ranging from standing up from a sitting position to standing on one leg. The subject is scored 0-4 points for each, reflecting their ability to perform the tasks independently, and the final measure is the sum of these. The BBS was shown by the authors to have excellent inter-rater and intra-rater reliability and to be internally consistent. It correlates satisfactorily with laboratory measures, such as postural sway, and was found to be a psychometrically sound measure of balance impairment for use in post-stroke assessment. The BBS also has have floor and ceiling effects, and it has been recommended to combine it with other balance measures [16]. Brassard et al. presented works towards an automatic, gamified version of the BBS [18].

Standard balance deficit test (SBDT) Tasks

- Standing on two legs with eyes open
- Standing on two legs with eyes closed
- Standing on one leg with eyes open
- Standing on one leg with eyes closed
- Eight tandem steps with eyes open
- Standing on two legs with eyes open on a foam support surface
- Standing on two legs with eyes closed on a foam support surface
- Standing on one leg on a foam support surface
- Eight tandem steps with eyes open on a foam support surface

- Walking 3 m with eyes open
- Walking 3 m with eyes open while rotating head
- Walking 3 m with eyes open while vertically pitching the head in rythm
- Walking 3 m forward with eyes closed
- Walking over four barriers

Quantitative Measurements

Force platforms (see Section 2.5.3) remain one of the most commonly used tools to measure balance in labs [12], [19]. For standing balance, the COP trajectory and its length are interesting metrics, that have recently been proposed to be alternatively measured with body-worn motion sensors [19]–[21].

Many of the tests we described in the last paragraphs have specific shortcomings [22]. The company APDM recently introduced the gait and posture mobility lab system[1] that aims at instrumenting these standard tests with a wireless, wearable sensor system based on several synchronized MARG sensors **MartinaManciniLaurieKing2013**. Their software suite provides modules for Timed Up and Go, Stand and Walk, and Instrumented Walk analysis. The advantage of using such a system is that it can compute detailed measures; for the Timed Up and Go test for example, it computes: cadence, gait cycle/double support/swing/stance time, stride length L/R, stride velocity, range of motion and velocity for arm, trunk and shank; turning duration, velocity and time and number of steps; stand-to-sit and sit-to-stand duration. The system can be worn over long periods of time, buffering the measurements for days and offloading them to the base station once within communication reach. This enables long-term gait analysis.

- Nice mobile posturography analysis of how age affects sway in different (standing/moving) conditions [23]

[1] `http://apdm.com/gait-and-posture/Mobility-Lab/`

- [24] video recording, Tandem Romberg position (w/ / w/o eyes closed), "For each subject, the effect of the use of tactile feedback was quantitatively analyzed using Standard Deviation and the number of measurements recorded outside the predefined boundary – Fraction Overshooting Threshold (FOT)."
- [25] (g)SBDT, measure sway before and after intervention

3.1.3 Treatment [5%]

Excercise Progression [26]

After Stroke [27]

Perturbation Techniques: Dynamic postural control can be measured by moving the base of support [12], [28] or the center of mass relative to each other, for example by pushing an individual, wobble boards, or actuated platforms.

3.1.4 Biofeedback for Balance [30%]

This should arch over the parameters commonly sensed, mappings of the feedback modalities, and in case of haptics places.

[5] [29]

Laboratory-based applications of biofeedback have been successful in improving balance in a number of populations [11], including healthy younger and older adults [30], individuals with vestibular loss [31]–[34], and post-stroke patients [35], [36].

The choice of the target modality depends on the application [37]. For sensorimotor relearning, the three major modalities that have been demonstrated are visual feedback [38], [39], auditory feedback [40], [41], or sonification [42], [43] (defined as the systematic non-verbal auditory representation of data [44]), and tactile feedback [33], [45]. For balance-related applications visual biofeedback of the body's center of pressure (COP) is most common, mapping the movement of the COP onto the motion of a virtual object such as a cursor [46].

Vibrotactile Balance Belts

Navigation, many (van Erp, blind)

Existing Balance Belts

EAI https://www.eaiinfo.com/sensory-kinetics/ Repulsive Cueing, Amount of lean will modulate pulse rate. SK Steady Stand provides tactile, visual and auditory feedback, showing COP, feedback limits and tactor activation map. Gamified, “roll to goal” patient uses their body motion to move a ball towards target over and undulating terrain.

Visual Feedback

just briefly give an overview

Auditory Feedback

Dozza M, Chiari L, Horak FB. Audio-biofeedback improves balance in patients with bilateral vestibular loss. Arch Phys Med Rehabil. 2005;86(7):1401–1403.

Provide multivariate auditory feedback during slackline stance. Anlauff, Fung, Cooperstock: Augmented Feedback for Learning Single-Legged

We developed an initial prototype with auditory feedback, described in Section A.

Vibrotactile Feedback

Prior literature has explored several modalities for giving feedback on posture, including auditory, haptic and electro-tactile [31] methods.

Haptics as a communication channel is uniquely beneficial for biofeedback as it can provide localized feedback [47]

Vibrotactile stimulation can also provoke involuntary reactions potentially beneficial for biofeedback [48]. Although the amount of information the tactile sense can resolve is less than for the auditory or visual sense [49],

Furthermore, tactile displays can be private, that is only felt by the user, and they are least likely to interfere with vision or hearing, which is especially important in populations with impaired vision or hearing and to support tasks in which the user relies on one of them [50].

Finally, touch is bidirectional in that it both supports sensing and acting on the environment [51](ToDo: *move?*).

These factors hopefully will allow our system to be used in more circumstances, for example while listening to music or talking. In this thesis, we will focus on tactile information delivery.

[52] feelSpace belt for sensory substitution

[24] uses Euler angles by Microstrain's Inertia Link (100 Hz), thresholding (2 degrees peak-to-peak, 3+ levels), two tactors

[53], van erp

[54] is the original one: "We have determined that the balance prosthesis should estimate the vertical within 0.1-1° to be useful to a patient [17], [35]. The estimated tilt error depends upon the instrumentation and algorithms." [55] conrad wall, the belt master, needs to be cited obviously; this one decribes the progression from kilogram-heavy hw to production-grade implementation: "The HG1920 IMU is a tactical grade instrument (Gyro bias in the 1 to 10°/hr range, and accelerometer gain sensitivity in the 0.001g to 0.005g range) 11. The drift in the tilt estimation algorithm was of the order of 0.1° over 10 hours. The IMU cost was $35,000." (ToDo: *compare w/ state of the art*) [34] another study of it

[56] et al.

[57] systems paper on a realworld belt

[45] similar evaluation on wobble board

[58] different encodings, acc + gyro

[25] vertiguart-rt: "determines continuously the coriolis force during body movements in pitch and roll by inbuilt gyroscopes and compares those values with individual preset thresholds for stimulator activation in specific directions."

(ToDo: *Sienko/Bao, what came after*)

3.2 HiFi Vibrotactile Belt mkI

3.2.1 Design Goals for a Haptic Balance Feedback Belt

The exploration on auditory posture feedback using a vision-based approach raised several problems that may impact its usefulness as a real-world in-home rehabilitation support system. Using a stationary, optical posture sensor requires stationary cameras and high-performance computing, both of which are usually stationary and may limit the opportunities for training to the specific spot it is set up on. Instead, we chose to use wearable motion sensors to measure postural parameters, such as trunk tilt. As discussed in (ToDo: *ref lit review section*), other have demonstrated that their measurements are appropriately accurate for providing balance feedback.

On the feedback side, we chose vibrotactile stimulation on the waist instead of auditory feedback. While able to deliver feedback on more than one postural parameter, relying on the auditory sense may be problematic if the users want to listen to music during exercise, communicate by speech or have impaired hearing. Such haptic augmented feedback also has the advantage of providing localized stimuli and thus being able to guide attention to specific body parts.

We investigate systems built with two types state-of-the art voice-coil tactors (ToDo: *line this up with what we defined in "background"*), one highly expressive that may allow us to render more information to the user through a single tactile channel, one type highly integrated and low in cost that we believe that these will become the norm in highly integrated, cost-effective haptics in the coming years.

Here we discuss the evolution of our highly expressive belt prototype design in three two interations, and one design for a highly integrated/compact design. These explorations form the basis for the modular toolkit described in Chapter 4, and in Chapter D we describe a subsequent design iteration for both designs based on the toolkit, along with proposed study design for their evaluation.

3.2.2 Hardware / Apparatus

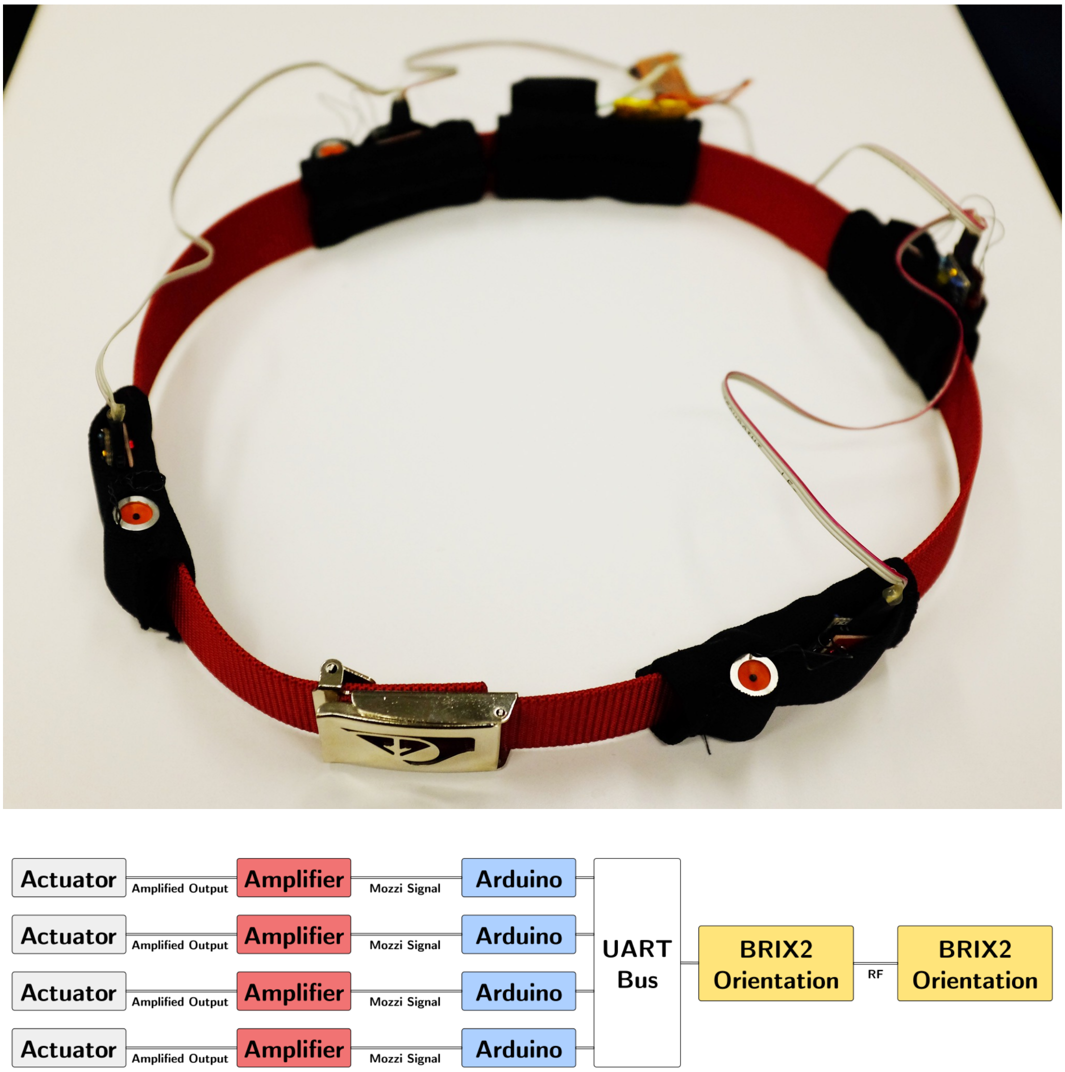


Figure 3.1: Haptic Belt mkI and System Architecture

Our first prototype was built around the Haptuator[2] voice coil tactor by Yao and Hayward [59], and ungrounded moving magnet voice-coil type linear actuators that allow arbitrary, potentially rich, waveforms to be rendered. (ToDo: *fix tense: present or past?*) Our initial prototype vibrotactile belt, shown in Fig. 3.1, is a wearable, embedded system capable of rendering haptic

[2] http://www.tactilelabs.com/products/haptics/haptuator/

feedback on trunk orientation without external system components. Location and duration of vibrotactile stimuli were found to be most promising for encoding information in haptic displays [51]. Placing the actuators around the waist allows us to give feedback physically close to the center of mass of the user, which we intend to guide. It was implemented as a proof-of-principle for a self-contained feedback system, and no external computer is involved in the synthesis of the feedback signal. Instead, the feedback computation and synthesis is performed by microcontrollers in the belt.

Our system is based on two $BRIX_2$ units in combination with four custom embedded synthesis units. $BRIX_2$ is a modular wireless sensor and microcontroller platform designed to provide building blocks for the development of body-worn wireless sensor networks and is described in Detail in Section ??. We used two base module and its Invensense MPU-9150 motion sensor chip, which incorporates three-axis accelerometer, gyrometer, and magnetometer sensors to measure trunk tilt. An onboard low-power wireless interface enables communication among several $BRIX_2$ units. Figure 3.2 shows one base module, with and without casing.

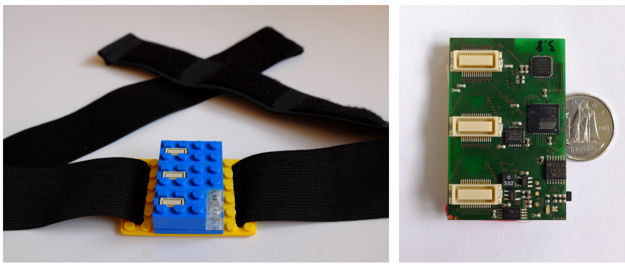


Figure 3.2: $BRIX_2$ Module on custom cheststrap and without case with 10 cents coin for size reference.

One $BRIX_2$ integrated into the belt serves as a central controller of the system, the other one provides sensor data. Our belt incorporates four Haptuators, placed at 0, 90, 180 and 270 degrees, corresponding exactly to forward, right, backward and left directions. This alignment should be optimal for giving directional feedback to the user, but unfortunately precludes us from measuring the orientation at these locations, both due to space constraints and possible interference of the Haptuator's strong magnets with the magnetometer. Thus, a second $BRIX_2$ worn on a chest strap senses the orientation of the user's body, which is communicated to the central controller via the wireless (RF) interface. The central $BRIX_2$ unit computes the feedback parameter mapping from these motion sensor readings and controls the synthesis parameters of the actuator units in the belt through a shared serial link.

Each of the four actuator units combines a Haptuator, an amplifier (Texas Instruments TPA2005D1, 1.4 Watt) and an Arduino Mini providing audio synthesis through the Mozzi library [60]. The $BRIX_2$, actuators and support electronics, including the battery power supply, are located in pouches made out of elastic and cotton fabric. Respectively, elastic band, Velcro hook and loop fabric fasteners and a Lego mounting plate form the cheststrap.

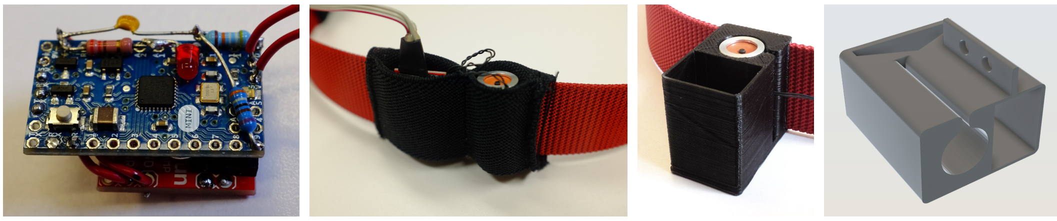


Figure 3.3: Arduino Mini and amplifier board, such an embedded synthesis unit together with a Haptuator in the initial belt pouch, and two subsequent iterations of the belt attachment.

As initial stimulus we chose a 250 Hertz sine wave, a frequency believed to provide maximum stimulation [61], with its amplitude reflecting the amount on the tilt of the chest accelerometer. Leaning the trunk off-axis will increase the vibration amplitude on the side of the tilt direction, guiding the user into a more upright position. "Pushing", instead of "attracting", the body into the correct position has been found to be more effective [62].

Software / Mapping

3.2.3 $BRIX_2$ Audio Extension

As a consequence of the work on the first-generation haptic belt, we developed the technological basis of the embedded synthesis units into an audio synthesis extension for the $BRIX_2$ platform. Existing approaches, such as the

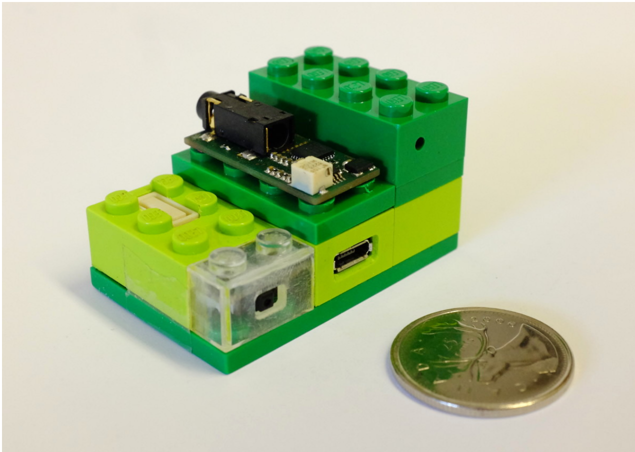
$BRIX_2$ vibrosound, use simple constant-frequency pulse-width modulation or wave-table lookup to generate continuous signals, such as sine or square waves. It enables on-body embedded synthesis by means of the Mozzi library [60] to generate audio or haptic stimuli with minimal latency, providing a set of sound synthesis building blocks more akin to desktop sound synthesis toolkits such as SuperCollider [63]. Up to three channels can be used per $BRIX_2$ module, and multiple $BRIX_2$ may be distributed over the body of the user. The extension is highly miniaturized and to our knowledge was the first time any of the modular prototyping systems was extended to provide high-fidelity synthesis capabilities and allows to directly drive an actuator or loudspeaker. Figure 3.4 (a) shows the extension on a base controller.

The extensions central component is a Atmel ATMega328P microcontroller dedicated to the synthesis, freeing the $BRIX_2$ microcontrollers to perform other tasks, such as processing sensor data. Communication with the base module, for example to set synthesis parameters, is through UART serial. The extension also includes the Texas Instruments TPA2005D1 1.5W D-Class amplifier, powered from the unregulated battery/USB voltage of the $BRIX_2$ base controller. Solder jumpers allow bypassing the amplifier for headphone output, and using the base controller for sound synthesis instead of the dedicated ATMega328P. The schematic of the extension is shown in Figure 3.4 (c), the board layout in Figure 3.4 (b). Programming of the ATMega328P on the extension is done by flashing a passthrough sketch on the $BRIX_2$ user controller and then programming it as another Arduino through the regular IDE. The audio extension is one of the most complex ones, and two iterations where designed and produced. The full documentation resides on the $BRIX_2$ website.[3] (ToDo: *applications?*)

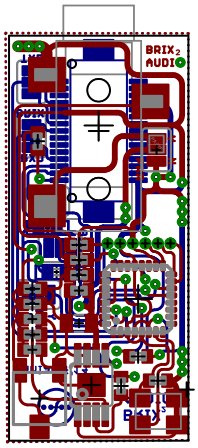
3.2.4 Discussion

- Actuator Mounts: The initial fabric pouches housing the embedded synthesis units proved to absorb a lot of the vibration energy and, together with the clothing users wore, did not lead to a good spatial separation of

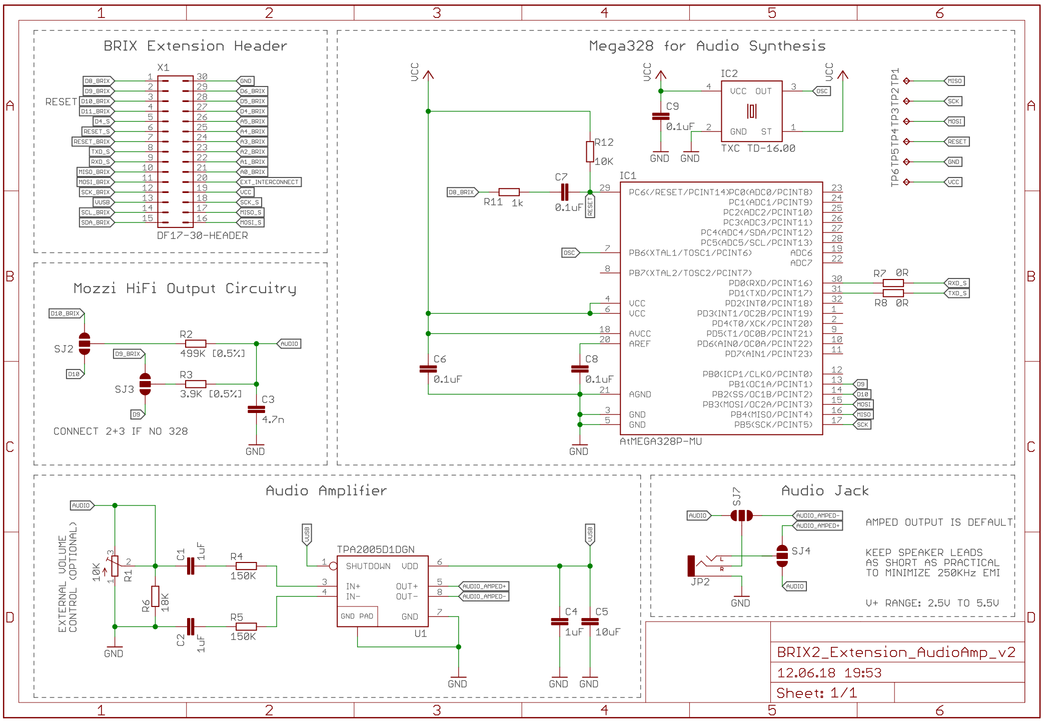
[3] https://opensource.cit-ec.de/projects/brix2/wiki/Audio_extension_new



(a) Extension with and without housing on top of base controller



(b) Board Layout



(c) Schematic

Figure 3.4: BRIX$_2$ audio extension

the stimuli. We evolved the mount through several iterations, eventually also rotating the actuator to be perpendicular to the body (ToDo: *fix reference to figure*) to improve energy transmission to the user.

- time is critical for effective stimulus design: flashing 5 microcontrollers is just not a great idea
- the MPUs sensor fusion was not good enough (drift), and it took up all the flash
- one-off prototype, fragile cabling
- rf interface not reliable, low latency but also low level
- Actuators bulky

3.3 HiFi Vibrotactile Belt mkII

3.3.1 Design Goals

3.3.2 4-Channel Embedded Audio Synthesis

One downside of the audio extension for use in the haptic belt prototype is that the $BRIX_2$ base controller only has three extension ports, while we use four actuators in our current design. Furthermore, the development cycles are slowed down by having to program multiple microcontrollers. The ATMega328 platform has been superceeded by arms, that feature a variety of benefits (flash, RAM, clock speed, peripherals) The flash memory, for example, was 90% full when including the firmware blob for the MPU-9120 MARG sensor fusion.

ARM-based microcontrollers are a 32-bit architecture that commonly has much larger memory and clock speeds. However, there is a large number of chips based on the architecture, ranging from low-power to the latest System-on-a-Chip units powering the latest smartphones. Traditionally, there is little support for end-users and open-hardware community by the chip manufacturers, and the diversity of the market meant their availability as development boards took quite some time.

A key characteristic in choosing a platform for rapid development is compatibility with the existing Arduino ecosystem. For this main reason, we chose the Teensy platform as the basis for our next generation embedded synthesis system.

Section 2.6.2

The Teensy Audio Adaptor[4] leverages this and provides two-channel output through one SGTL5000 stereo codec. The author provides a web-based configuration tool for the signal routing and an Arduino library library support to use the DSP instructions easily. As the K20 has supports two I2S channels, two stereo codecs can be connected, resulting in four channel audio output, two channel audio input.

The intent for this redesign was to use it for a next generation haptic shoe based on voice coil actuators, described in Section C.3. Thus, we based the first prototype on the proven design of the TT-200 shoe controller design, described in Section B.4.3. We kept the Bosch BNO-055 9-DoF motion sensor and the operational amplifiers used to condition the FSR pressure sensor signals described in Section B.4.3. Instead of the four LRA/ERM driver chips, we included the circuitry required for embedded four-channel synthesis based on two NXP SGTL5000 chips, sharing a common reference power supply. The module is self-operating, but has the footprint and extension header of the $BRIX_2$ system, so it could be also considered a four-channel $BRIX_2$ extension. Given the much higher processing power of the K20 MCU, the $BRIX_2$ system could also be considered an extension to the TT-300, providing access to it's RF communication and MPU-9250 motion sensor. Communication with $BRIX_2$ would either be through UART or I2C, and the $BRIX_2$ module would provide power in this scenario. Direct wireless communication is possible through a separate UART/I2C header meant to connect a BLE Nano or similar, as with the TT-200. The input of four channels of FSR force sensors and the audio ouput of the four signal/ground pairs is through MicroMatch headers. Solderjumpers allow choosing between line-level output and output through the SGTL5000 integrated headphone amplifier. Figure 3.5 show top and bottom of the TT-300 module, Figure 3.5 (c) details the schematic.

[4] https://www.pjrc.com/store/teensy3_audio.html

3.3.3 Lofelt L5 Tactors

- Tactile Labs Haptuators attractive for basic research due to ther linear frequency response, but high price makes practical prototypes very costly (250 USD per unit)
- Long-time durability was not great for some of the Haptuators,
- power of planar only sufficient for some applications, Concerns over durability when used in shoes, availability poor
- Talent behind them moved on to commercial ventures; Actronica and Lofelt
- Lofelt Sound Engine (despite being quiet in the original application) was engineered for their Basslet, a , watch-shaped "wearable subwoofer"(Add Citation: *urlhttps://lofelt.com/the-basslet*); low-pass at 250 Hz
- durable design with spring, industrially made
- can be reliably sourced, price point @ 20 USD for educational small quantities
- low profile, ideal for shoes, but for integration only, fragile
- two peaks @ 38 and 60 Hz (ToDo: *verify*), strong; see Fig. 3.6 3.6c

3.3.4 Amplifier Module

The original design intent was to be able to integrate it with these comparatively small actuators for a wrist-worn tactile display to communicate alarms in high attentional load environments (ToDo: *reference alarm systems applications*) and for in-and-around-the wrist effects (ToDo: *ref to ROI*), as well as in the third haptic shoe generation.(ToDo: *move this to a general system design thing at the beginning*) The actuator of choice at the time was the Tactile Labs Haptuator Planar, described in Section 2.4.3, which was the smallest commercially available high-fidelity tactor at the time of development. The first version of the amplifier module (AMP-100) uses an 1.5W D-class amplifier (Texas Instruments TPA2005D1). The module's footprint is 12 × 12 mm^2 matches the one of the Haptuator Planar to be able to integrate the two in a vertical stack, as illustrated in Figure 3.7 (a). The power pins are included on both sides of the PCB, shown in Figure 3.7 (b), to be able to daisy chain a power supply through multiple boards, such as in the shoe, reducing the amount of wires to run.

The result is a compact module that is much smaller than the ordinarily available breakout boards, such as the one by Sparkfun (ToDo: *ref?*). We 3D-printed a 1 inch strap-mount housing that holds a Lofelt tactor and the amplifier module, which can be seen in Figure 3.7 (b). It can be worn, for example, like a watch on the wrist. The demonstrator was used in a clinical in-situ study conducted by Lindenbaum et al. to investigate the delivery of vibrotactile stimuli to locked-in syndrom patients and their ability to trigger P300 event-related EEG potentials for a novel brain-computer interface.(ToDo: *cite*)

3.3.5 Audio Interconnect

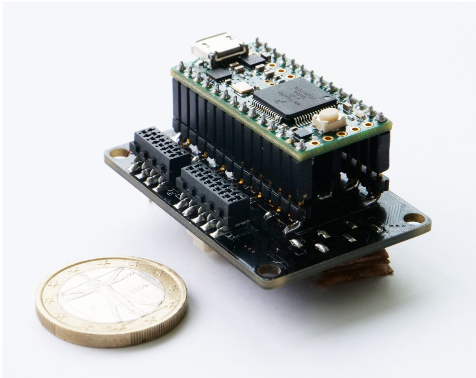
We need to be able to connect standard 1/8 inch phone jack audio sources to our system. This may be for prototyping, where it is much easier to synthesize the signals on a desktop computer, or because more computing resources than the Teensy provides are required. (ToDo: *put this into requirements much earlier*) As the distance between input and output can vary between applications, we chose to base this potentially multi-meter cabling stretch on RJ45 ethernet cabling. Figure 3.8 3.8a shows the ACL-100 (line/input side) and ACS-100 ("shoe"/placement side) audio connector boards, which connect out four channels (two stereo pairs) through a standard ethernet cable to the MicroMatch-8 connector on the placement side.

The MicroMatch-8 input on the placement, either coming the TT-300 main board or from another external source, has to be split into individual lines to distribute the audio signals to the amplifier modules. Two grounds are necessary as the audio ground may not be the same as the power ground; this is the case with the output of the SGTL5000 headphone amplifier for example. (ToDo: *no schematic explanation? too simple*) Figure 3.8 (b) shows the distributor with all five cables connected, and the schematic for reference.

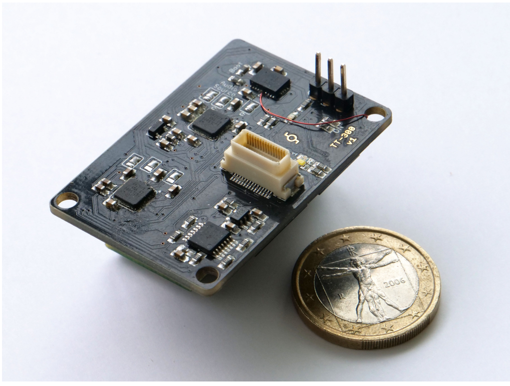
3.3.6 Power Module

This module is separated from the main unit to be able to place it together with the battery, which in the case of lithium polymer batteries may present a component that is safety critical and may need special housing. Furthermore, it isolates the switching regulator from the analog parts of the circuit.

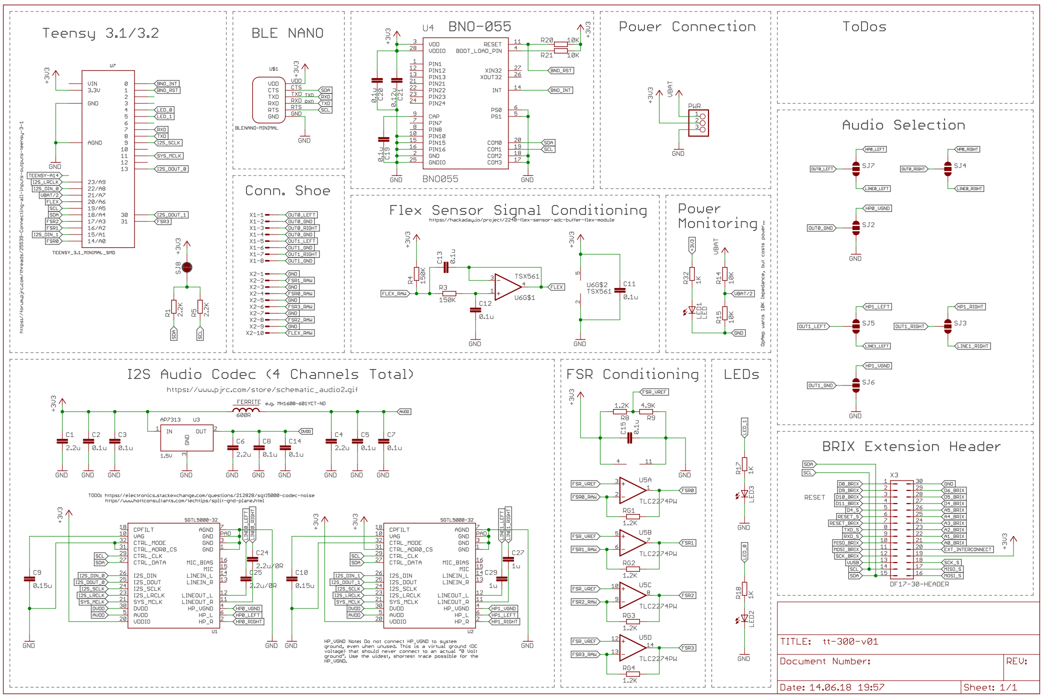
We use the proven combination of a Texas Instruments TPS63031 switching regulator (1A switch), TI TPS2115A power multiplexer and Microchip MCP73831OT lithium battery charger chips. The power module is able to provide up to (ToDo: *??*) amps of continous power at (ToDo: *x*) percent efficiency. The output of the module is a three-pin header (0.1 inch pitch) with +3.3V, the unregulated battery voltage, and ground. Batteries are connected through the common JST PH 2-Pin connector. Figure 3.9 shows the module and the schematic.



(a) Top View



(b) Bottom View

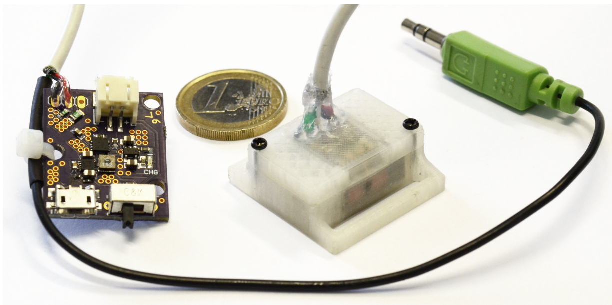


(c) Schematic

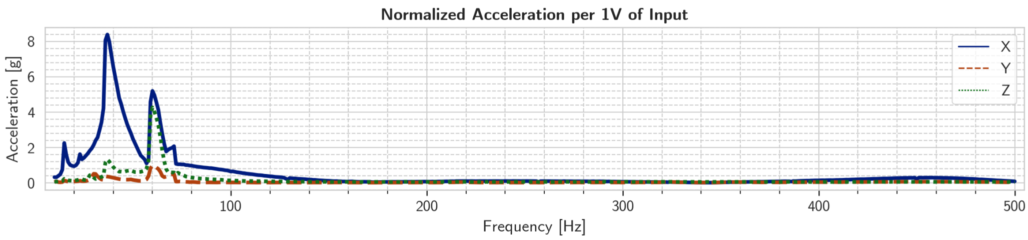
Figure 3.5: First version of audio synthesis / motion sensing module (TT-300)

(a) Lofelt L5 Tactor, unpackaged.

,

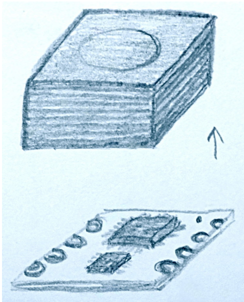


(b) Demonstrator: Amp-100 with with Lofelt L5 tactor in strap housing

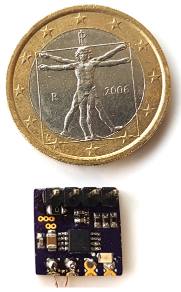


(c) Lofelt L5 Tactor frequency response. Test mass 100 g, 1 Hz Steps, 5 repetitions, light blue shows 95% confidence interval.

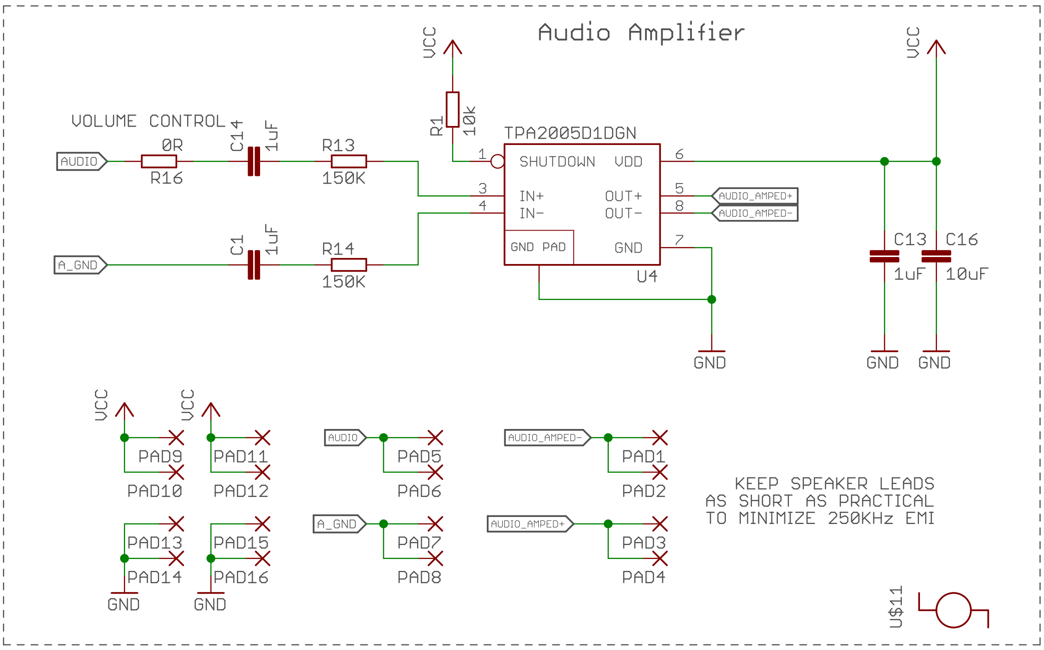
Figure 3.6: Lofelt L5 Tactor, packaged, and actuator frequency response.



(a) Actuator Shield Concept

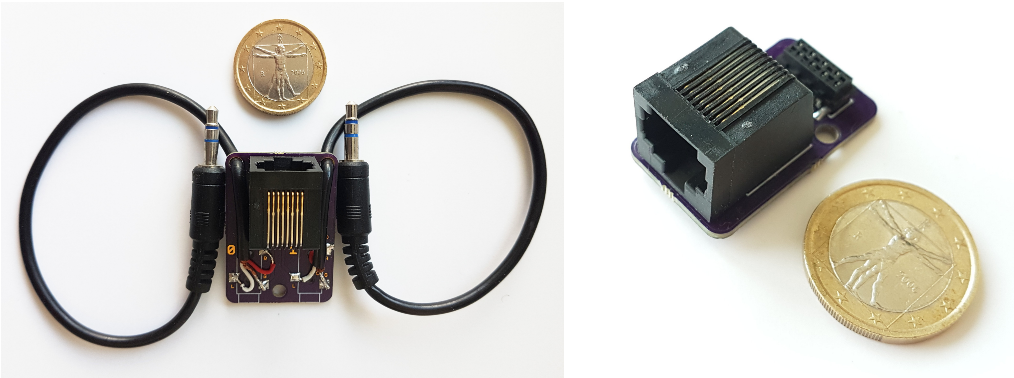


(b) PCB

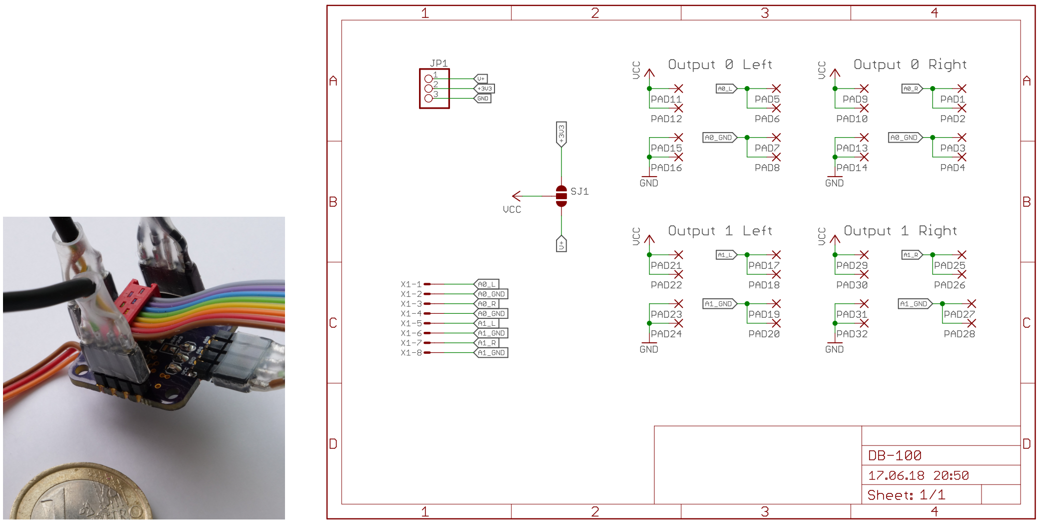


(c) Schematic

Figure 3.7: First version of amplifier module (AMP-100)

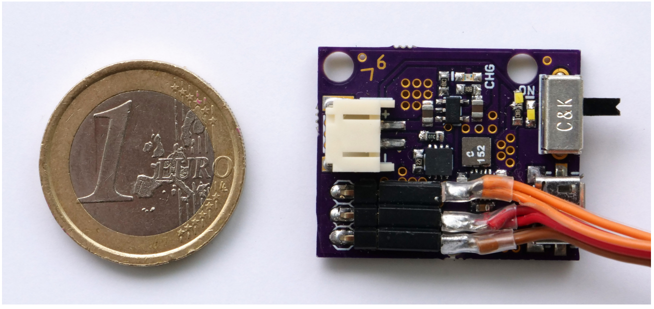


(a) Audio Connector Boards (ACL-100 & ACS-100)

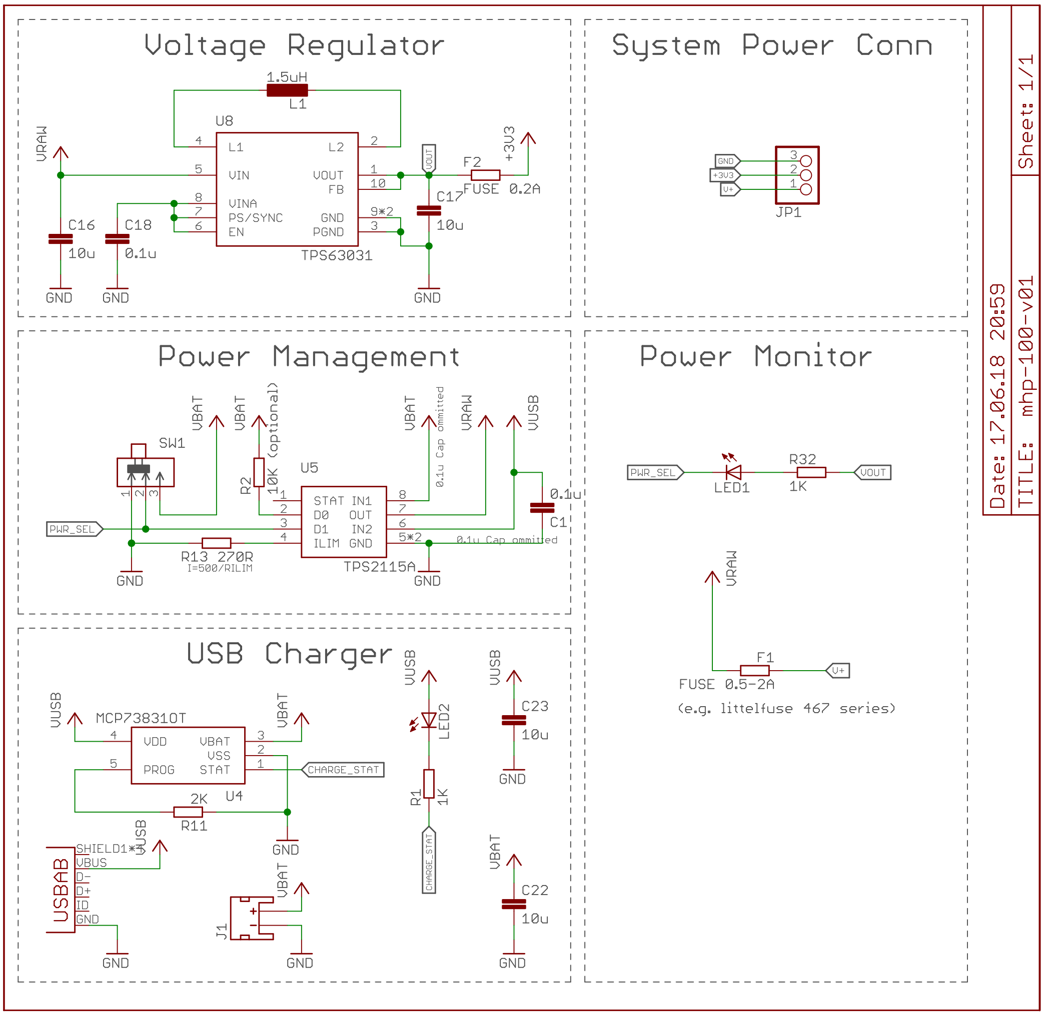


(b) Distributor Board (DB-100)

Figure 3.8: Audio Connector and Distributor Board

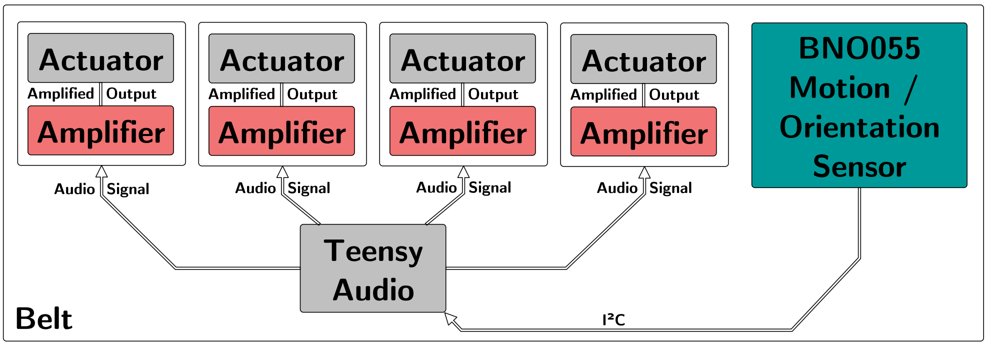


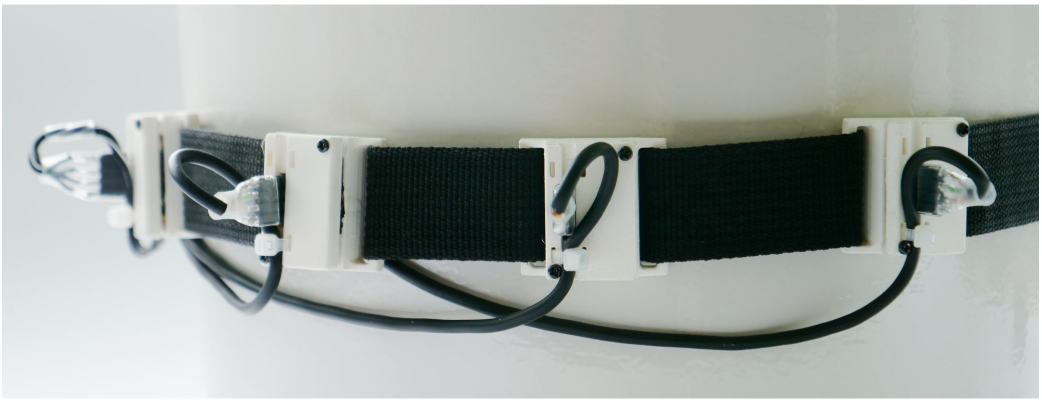
(a) Power module, first generation



(b) Schematic

Figure 3.9: First version of power (MHP-100)





3.3.7 Software / Mapping

3.3.8 Discussion

The audio synthesis and the sensing worked individually as expected, however, we ran into issues combining the readout of the BNO-055 motion sensor with the synthesis. After lengthy debugging, involving two different breadboard setups and communication with the Teensy creator[5], the problem seems to be the 11.3 MHz square wave for the I2S master clock (MCLK) that can radiate RF energy, to which the BNO0055 appears to be particularly sensitive. As this combination is the key for closed-loop feedback systems, the failure to operate in tandem meant we had to redesign the circuit to try and avoid the interference.

The module interconnect was hand-made with shrink tubing and standard 0.1 inch male/female pin headers. This was rapid to implement and allowed us

[5] https://forum.pjrc.com/threads/40635-I2S-audio-output-and-problems-w-I2C-IMU

to test the system without procuring specialized plugs, but is time-consuming and mechanically not as robust as a proper interconnect. On the amplifier boards, the vertical exit of the wire lead to cables protruding perpendicular to the actuator unit, which can be seen in Figure 3.6b.

The vertical height of the previous prototypes, resulting from the pinheaders used to connect TT-300 board and Teensy, made the system relatively bulky and hard to integrate.

good:

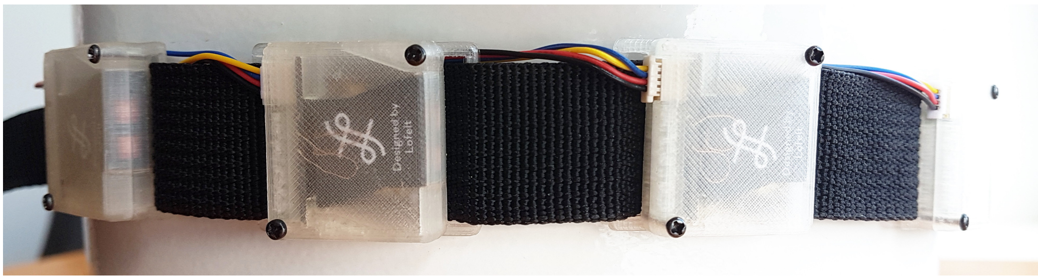
- proper 4ch embedded synthesis w/ teensy instead of 5 arduinos trying to do it poorly
- high-quality marg orientation sensor
- new actuators: cheaper, smaller, better power-to-weight ratio (probably)
- new modular achitecture
- also support for shoe controller function
- support for using rj45 cable for external signal synthesis

not so good

- cabling suboptimal: direction, cables had to be made by hand
- teensy controller bulky and difficult to solder
- rfi problem bno/teensy audio rendered motion sensor not working

3.4 HiFi Vibrotactile Belt mkIII

3.4.1 Hardware



good:

- serves as proof-of-concept rapid prototyping application for toolkit
- better amplifiers, smaller boards, better actuator housing
- base everything off JST SH connectors, reduce complexity of individual modules, no more connector/cable making!
- more compact, reliable teensy integration
- adaptors for tethered operation
- separate bno on module, resolve rfi problem (shielded cable now possible)
- separate analog circuitry on other system
- support arbitrary i2c sensors
- stretchy belt

3.5 Proposed Evaluation of Vibrotactile AF for Hip Rotation

- Feedback on balance is likely to be most effective when used over longer times and becomes part of non-conscious processes, e.g., during longitudinal studies such as Sienko's long-term balance trainer [2].(Add Citation: *biofeedback takes time?*)

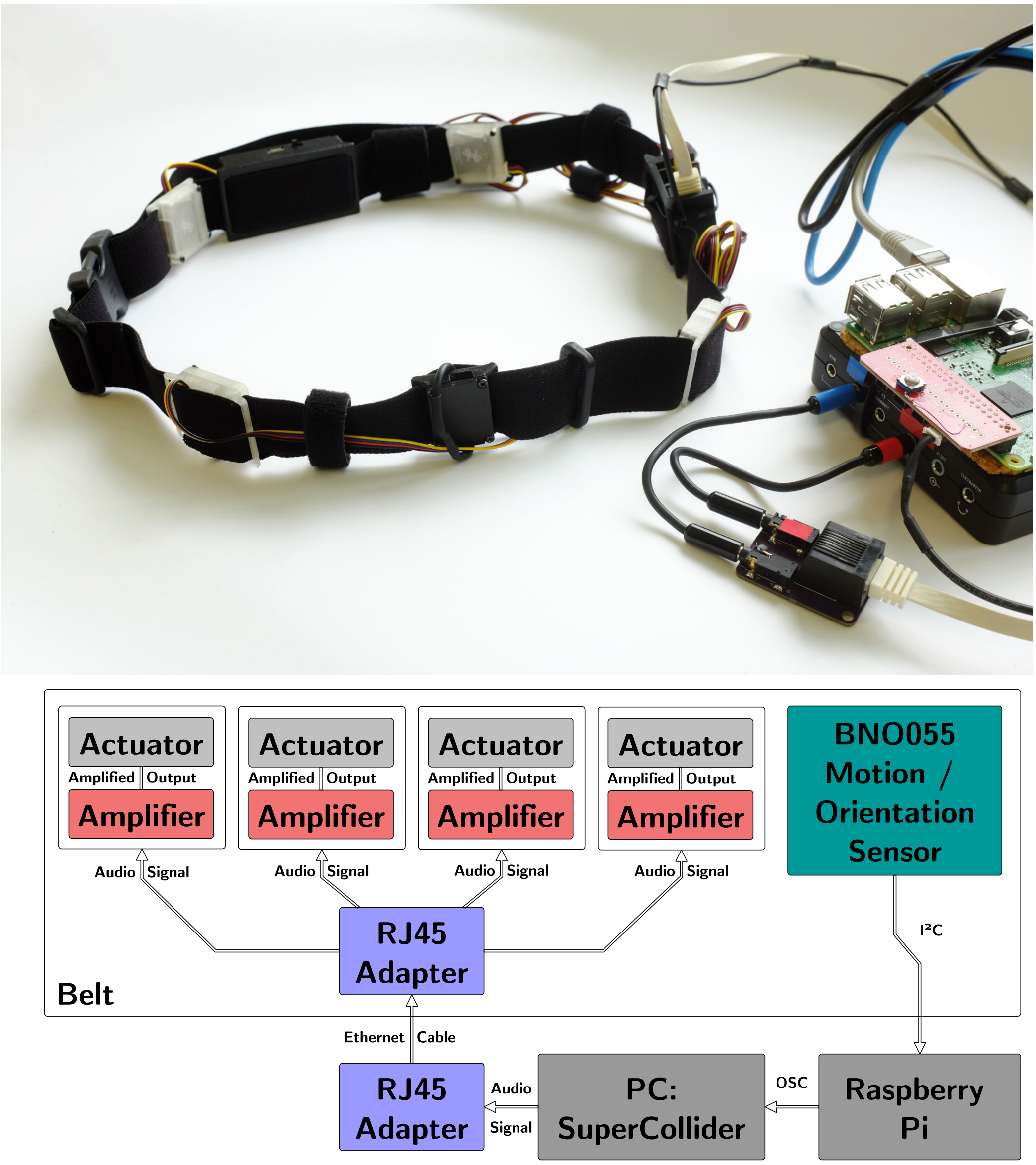


Figure 3.10: Vibrotactile Belt mkIII, tethered setup. Four Lofelt tactors with integrated amplifiers, one BNO055 motion/orientation sensor, power supply on belt. Signals supplied by computer (not shown) running SuperCollider through RJ45 cable (white/flat) on sensor data received through OSC from Raspberry Pi on the right that reads out the BNO055 sensor.

- This was confirmed during pilot test to replicate a setup similar to study we found that subjects found it hard to incorporate balance feedback during challenging conditions, such as single-legged stance, in addition to the intrinsic feedback.
- for a pilot study investigating balance feedback, testing different feedback mappings.
- estimate angular displacements (tilt angles) in anterior-posterior and medial-lateral direction through BN055
- sienko:

> The tactor activation signal was defined as the tilt angle plus one half times the tilt angular rate for Categories 1, 2, 4 and 5, and as the tilt angle for Category 3 exercises [24]. If the tactor activation signal exceeded a pre-set threshold [23, 38], the sensing unit sent audio output signals to the tactor bud accessory. The tactor bud accessory ana- lyzed these audio signals and activated the correspond- ing tactor to provide vibrotactile cues. At the end of each repetition, the trunk motion data, number of step-outs, and visual analog scale ratings were auto- matically uploaded to a secured server via Wi-Fi.

- We chose to validate the functionality of our system with a simpler excercise that allows to consciously integrate the feedback signal.

3.5.1 State of the Art

Research Questions

- Can we reproduce improved performance with univariate feedback?
- Can users distinguish the two types of vibrotactile stimuli on the waist?
- Is multivariate balance feedback more effective?

Pitfalls/Challenges

- mapping

- questionnaires
- balance metric?
- Sensor/Actuator Placement
- Calibration?
- proportional+derivative feedback?
- sensors good enough? Sway star: 0.002 deg/s drift (4deg/h/1deg/h), sd 0.02 deg/s in practice; +- 0.4 deg/s quiet sway; mems: 0.4 deg/s sd 0.4 deg/s drift mean; sienko happily uses iphone

Apparatus

3.6 Low Cost LRA Belt

- why? low-power, highly miniaturized modular kit for building something like a low cost belt
- how? leverage ble nano; add stacked shields
- good? headers hard to get, vertical height problematic: directly motivates cable-based interconnect; using prototyping technology not supported by larger companies such as adafruit/sparkfun can easily get you deprecated: redbearlabs acquired by IOT company, products cancelled

3.6.1 Hardware / Apparatus

The nRF5x Redbear BLE Nano used for BLE communication in the shoe project actually is a quite capable microcontroller (See Section 2.6.1) that may be sufficient for . The TI DRV2605L driver chips run over I²C, they can be directly controlled through from the BLE Nano MUC. It is designed for low-power (e.g., running of a coin cell) and LRAs are also optimized to be power-efficient, thus the combination of these provides ideal basis for a system that supports extended runtimes for in-situ application of a comparatively small battery. Here, we discuss a prototype system for such an application that is a stack of shields to the BLE Nano.

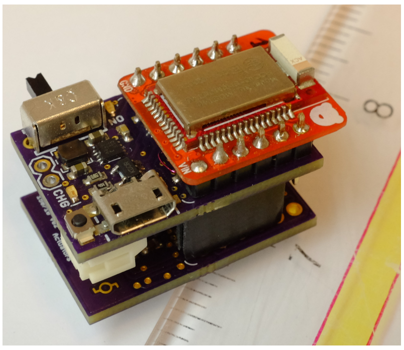
The base shield is shown in Figure 3.11 (a). It provides headers for the BLE Nano, the proven power circuitry (LiPo charger, regulator, power mux), and a Bosch BNO055 9-DoF motion sensor with integrated sensor fusion. It includes a single DRV2605L module to drive one LRA/ERM directly. This combination would provide a direct replacement, for example, the Pebble smartwatch used in studies such as [64], while providing much more control over the stimulus and sensor readings. the modules are similar in size to a watch with a 30×20 mm^2 footprint. The schematic is show in Figure 3.11 (c).

For four or eight channels of output, one or two four channels LRA/ERM driver boards can be stacked underneath the base shield, as shown in Figure 3.11 (b). , a 4-channel I^2C multiplexer (PCA9546) and four DRV2605L drivers are the main components; the schematic is shown in Fig. 3.11 3.11c

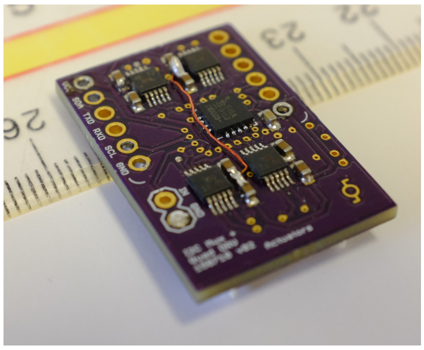
The actuators connect through a Molex Picoblade connector with crimp-style cables and 1.25 mm pitch. As crimping the cables by hand is feasible with the right technique[6] and tools[7] but fairly difficult, we opted for pre-made cable assemblies from Molex that just need to be mated with the tactors.

This design forms the basis for a module of the final toolkit, described in Section 4.2.4.

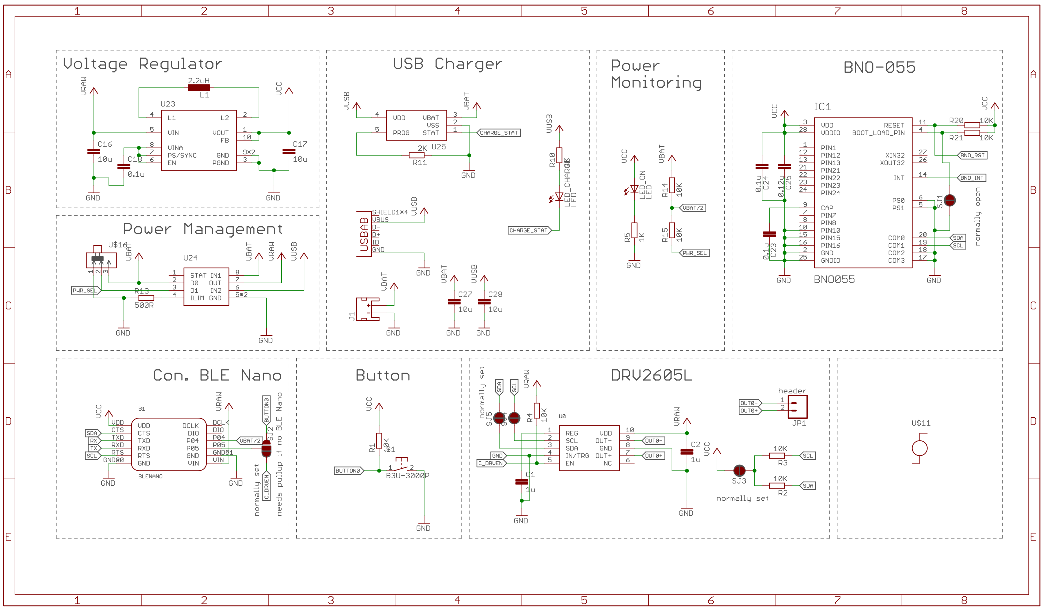
[6]https://www.youtube.com/watch?v=Ta55NTSBLN0
[7]http://www.engineer.jp/en/products/pa09e.html



(a) BLE Nano on base shield stacked on quad driver shield



(b) BLE Nano quad driver shield, top, ruler (inches) for size reference.



(c) Base shield schematic

Figure 3.11: BLE Nano shield system

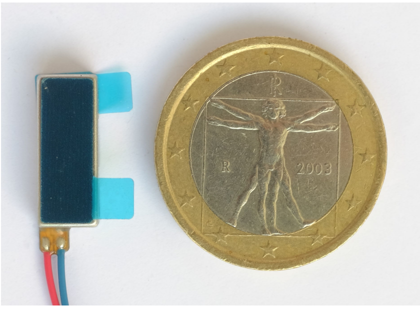
3.6.2 Software / Trainer Application

As a proof-of-principle for the rapid development of a vibrotactile balance feedback system, we consider an LRA-based belt. → BSC-Thesis by Johannes Schlueter based on my software and hardware → Redone w/ Feather - became brix5

- WebBluetooth interface
- good example for a student project

LRAs

We chose to use the largest LRAs readily at the time of development, the Jin Long (ToDo: ??). Its shown in Fig. 3.12, encased in a variant of the parametric belt mount and without.



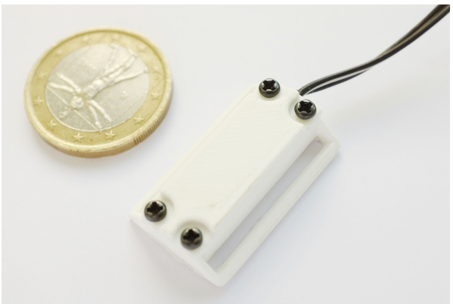


Figure 3.12: Linear Resonant Actuator, with and without housing

3.6.3 Discussion

3.7 Conclusion

Wrap this up, details should be in the sections, consequence in the modular chapter.

Lessons Learned:

- Good actuator-body (de-)coupling is critical
- Adjustability vs. Robustness
- Modular systems are key for rapid development: student proposals could be executed in short time if hw development would not have to take place each time.

3.8 Publications and Disclosures

https://ieeexplore.ieee.org/abstract/document/6954321/

Anlauff: Wearable Computing for Ubiquitous Balance Assessment and Training, IEEE HAVE'14 Doctoral Consortium

Anlauff, Fung, Cooperstock: Modular Haptic Belt for Augmented Balance Feedback, ISPGR'15 (Poster)

Coupling Prediction ROI Anlauff, Blum, Cooperstock: Inferring haptic actuator coupling from actuator current draw. 2016

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Belt mkI

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LRA-Belt

This project was executed as a bachelors thesis by Johannes Schlueter, I provided hardware and software for the embedded part, and the shoe controller applicatin as starting point for the mobile application.

References for Chapter 3

[1] K. H. Sienko, S. L. Whitney, W. J. Carender, and C. I. Wall, "The Role of Sensory Augmentation for People with Vestibular Deficits: Real-Time Balance Aid and/or Rehabilitation Device?" *Journal of Vestibular Research*, vol. 27, no. 1, pp. 63–76, Apr. 2017, ISSN: 09574271, 18786464. DOI: 10/f9zw8f.

[2] T. Bao, W. J. Carender, C. Kinnaird, V. J. Barone, G. Peethambaran, S. L. Whitney, M. Kabeto, R. D. Seidler, and K. H. Sienko, "Effects of Long-Term Balance Training with Vibrotactile Sensory Augmentation among Community-Dwelling Healthy Older Adults: A Randomized Preliminary Study," *Journal of NeuroEngineering and Rehabilitation*, vol. 15, no. 1, 2018, ISSN: 1743-0003. DOI: 10/gdwstf.

[3] F. B. Horak, "Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls?" *Age and ageing*, vol. 35 Suppl 2, pp. ii7–ii11, Sep. 2006, ISSN: 0002-0729. DOI: 10.1093/ageing/afl077. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16926210.

[4] B. Lauber and M. Keller, "Improving motor performance: Selected aspects of augmented feedback in exercise and health," *European Journal of Sport Science*, Sep. 2012, ISSN: 1746-1391. DOI: 10.1080/17461391.2012.725104. [Online]. Available: http://www.tandfonline.com/doi/abs/10.1080/17461391.2012.725104.

[5] A. Zijlstra, M. Mancini, L. Chiari, and W. Zijlstra, "Biofeedback for training balance and mobility tasks in older populations: a systematic review.," *Journal of neuroengineering and rehabilitation*, vol. 7, no. 1, p. 58, Jan. 2010, ISSN: 1743-0003. DOI: 10.1186/1743-0003-7-58. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3019192&tool=pmcentrez&rendertype=abstract.

[6] M. Dozza, "Biofeedback systems for human postural control," Ph.D. Thesis, Bologna University, 2007. [Online]. Available: http://amsdottorato.cib.unibo.it/391/1/Marco_Dozza_-_Thesis.pdf.

[7] A. Olofsson, S.-O. Andersson, and B. Carlberg, "'If only I manage to get home I'll get better' - Interviews with stroke patients after emergency stay in hospital on their experiences and needs," *Clinical rehabilitation*, vol. 19, no. 4, pp. 433–440, Jun. 2005, ISSN: 0269-2155. DOI: 10.1191/0269215505cr788oa. [Online]. Available: http://cre.sagepub.com/cgi/doi/10.1191/0269215505cr788oa.

[8] M. V. Radomski, "More than good intentions: advancing adherence to therapy recommendations.," *The American journal of occupational therapy : official publication of the American Occupational Therapy Association*, vol. 65, no. 4, pp. 471–7, 2006, ISSN: 0272-9490. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/21834463.

[9] K. Andrews and J. Stewart, "Stroke recovery: He can but does he?" *Rheumatology and rehabilitation*, vol. 18, no. 1, pp. 43–8, Feb. 1979, ISSN: 0300-3396. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/424667.

[10] A. L. Betker and T. Szturm, "Balance Costs of Standing on Compliant Surfaces in Healthy Subjects," pp. 1–18, 2004.

[11] T. E. Howe, L. Rochester, A. Jackson, P. M. H. Banks, and V. A. Blair, "Exercise for improving balance in older people," *The Cochrane database of systematic reviews*, no. 4, p. CD004963, 2007, ISSN: 1469-493X. DOI: 10.1002/14651858.cd004963.pub2. [Online]. Available: http://dx.doi.org/10.1002/14651858.CD004963.pub2.

[12] F. M. Collen, "The measurement of standing balance after stroke," *Physiotherapy Theory and Practice*, vol. 11, no. 2, pp. 109–118, Jan. 1995, ISSN: 0959-3985. DOI: 10.3109/09593989509022407. [Online]. Available: http://informahealthcare.com/doi/abs/10.3109/09593989509022407.

[13] B. L. Riemann, J. B. Myers, and S. M. Lephart, "Sensorimotor system measurement techniques.," *Journal of athletic training*, vol. 37, no. 1, pp. 85–98, Jan. 2002, ISSN: 1938-162X. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=164313&tool=pmcentrez&rendertype=abstract.

[14] K. C. Hayes, "Falls and instability in the elderly," *Physiotherapy Canada*, vol. 42, pp. 59–60, 1990.

[15] A. R. Fugl-Meyer, L. Jääskö, I. Leyman, S. Olsson, and S. Steglind, "The post-stroke hemiplegic patient. 1. A method for evaluation of physical performance.," *Scandinavian journal of rehabilitation medicine*, vol. 7, no. 1, pp. 13–31, 1974.

[16] L. Blum and N. Korner-Bitensky, "Usefulness of the Berg Balance Scale in Stroke Rehabilitation: A Systematic Review," English, *Physical Therapy*, vol. 88, no. 5, pp. 559–566, May 2008, ISSN: 00319023. [Online]. Available: http://search.proquest.com/docview/223115945?accountid=12339.

[17] K. Berg, "Measuring balance in the elderly: preliminary development of an instrument," *Physiotherapy Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[18] S. Brassard, M. J.-D. Otis, A. Poirier, and B.-A. J. Menelas, "Towards an Automatic Version of the Berg Balance Scale Test Through a Serious Game," in *Proceedings of the Second ACM Workshop on Mobile Systems, Applications, and Services for HealthCare*, ser. mHealthSys '12, New York, NY, USA: ACM, 2012, 5:1–5:6, ISBN: 978-1-4503-1764-1. DOI: 10.1145/2396276.2396282. [Online]. Available: http://dl.acm.org/citation.cfm?id=2396282%20http://doi.acm.org/10.1145/2396276.2396282.

[19] C. Seimetz, D. Tan, R. Katayama, and T. Lockhart, "A comparison between methods of measuring postural stability: force plates versus accelerometers," *Biomedical sciences instrumentation*, vol. 48, no. March 2012, p. 386, 2012.

[20] A. L. Betker, T. Szturm, and Z. M. K. Moussavi, "Estimation of 2-D center of mass movement during trunk flexion-extension movements using body accelerations.," *IEEE transactions on neural systems and rehabilitation engineering*, vol. 17, no. 6, pp. 553–9, Dec. 2009, ISSN: 1558-0210. DOI: 10.1109/TNSRE.2009.2032620. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/19775986.

[21] A. L. Betker, Z. M. K. Moussavi, and T. Szturm, "Ambulatory center of mass prediction using body accelerations and center of foot pressure.," *IEEE transactions on bio-medical engineering*, vol. 55, no. 11, pp. 2491–8, Nov. 2008, ISSN: 1558-2531. DOI: 10.1109/TBME.2008.2001127. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/18990618.

[22] M. Mancini and F. B. Horak, "The relevance of clinical balance assessment tools to differentiate balance deficits.," *European journal of physical and rehabilitation medicine*, vol. 46, no. 2, pp. 239–48, Jun. 2010, ISSN: 1973-9095. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3033730&tool=pmcentrez&rendertype=abstract.

[23] D. Basta, M. Rossi-Izquierdo, A. Soto-Varela, and A. Ernst, "Mobile posturography: posturographic analysis of daily-life mobility.," *Otology & neurotology: Official publication of the American Otological Society, American Neurotology Society [and] European Academy of Otology and Neurotology*, vol. 34, no. 2, pp. 288–97, 2013, ISSN: 1537-4505. DOI: 10.1097/MAO.0b013e318277a29b. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/23151777.

[24] A. U. Alahakone and S. M. N. A. Senanayake, "A combination of inertial sensors and vibrotactile feedback for balance improvements in therapeutic applications," *2009 Innovative Technologies in Intelligent Systems and Industrial Applications, CITISIA 2009*, no. July, pp. 5–10, 2009. DOI: 10.1109/CITISIA.2009.5224252.

[25] M. Rossi-Izquierdo, A. Ernst, A. Soto-Varela, S. Santos-Pérez, A. Faraldo-García, A. Sesar-Ignacio, and D. Basta, "Vibrotactile neurofeedback balance training in patients with Parkinson's disease: reducing the number of falls.," *Gait & posture*, vol. 37, no. 2, pp. 195–200, Feb. 2013, ISSN: 1879-2219. DOI: 10.1016/j.gaitpost.2012.07.002. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/22841586.

[26] B. N. Klatt, W. J. Carender, C. C. Lin, S. F. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, "A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular

Disorders," *Physical medicine and rehabilitation international*, vol. 2, no. 4, p. 1044, Apr. 2015, ISSN: 2045-2322. [Online]. Available: http://www.ncbi.nlm.nih.gov/pmc/articles/PMC4968039/.

[27] P. Langhorne, F. Coupar, and A. Pollock, "Motor recovery after stroke: a systematic review.," *Lancet neurology*, vol. 8, no. 8, pp. 741–54, Aug. 2009, ISSN: 1474-4422. DOI: 10.1016/S1474-4422(09)70150-4. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/19608100.

[28] F. B. Horak, S. M. Henry, and A. Shumway-Cook, "Postural perturbations: new insights for treatment of balance disorders.," en, *Physical therapy*, vol. 77, no. 5, pp. 517–33, May 1997, ISSN: 0031-9023. [Online]. Available: http://ptjournal.apta.org/content/77/5/517.abstract.

[29] O. M. Giggins, U. M. Persson, and B. Caulfield, "Biofeedback in rehabilitation," *Journal of neuroengineering and rehabilitation*, vol. 10, no. 1, p. 60, Jan. 2013, ISSN: 1743-0003. DOI: 10.1186/1743-0003-10-60. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3687555&tool=pmcentrez&rendertype=abstract.

[30] L. L. Verhoeff, C. G. C. Horlings, L. J. F. Janssen, S. A. Bridenbaugh, and J. H. J. Allum, "Effects of biofeedback on trunk sway during dual tasking in the healthy young and elderly," *Gait & Posture*, vol. 30, no. 1, pp. 76–81, Jul. 2009, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2009.03.002. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636209000733.

[31] M. E. Tyler, Y. Danilov, and P. Bach-Y-Rita, "Closing an open-loop control system: vestibular substitution through the tongue," *Journal of integrative neuroscience*, vol. 2, no. 2, pp. 159–164, 2003. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/15011268.

[32] M. Dozza, F. B. Horak, and L. Chiari, "Auditory biofeedback substitutes for loss of sensory information in maintaining stance.," *Experimental brain research.*, vol. 178, no. 1, pp. 37–48, Mar. 2007, ISSN: 0014-4819. DOI: 10.1007/s00221-006-0709-y. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/17021893.

[33] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. I. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *Journal of Vestibular Research*, vol. 18, no. 5, pp. 273–285, 2008.

[34] D. Basta and A. Ernst, “Rehabilitation von Gleichgewichtsstörungen mithilfe eines vibrotaktilen Neurofeedbacktrainings,” *HNO kompakt*, vol. 19, no. 3, pp. 1–3, 2011.

[35] M. B. Badke, J. Sherman, P. Boyne, S. Page, and K. Dunning, “Tongue-based biofeedback for balance in stroke: results of an 8-week pilot study,” *Archives of physical medicine and rehabilitation*, vol. 92, no. 9, pp. 1364–70, Sep. 2011, ISSN: 1532-821X. [Online]. Available: `http://dx.doi.org/10.1016/j.apmr.2011.03.030`.

[36] S. H. Hong, S. Im, and G.-Y. Park, “The effects of visual and haptic vertical stimulation on standing balance in stroke patients.,” *Annals of rehabilitation medicine*, vol. 37, no. 6, pp. 862–70, Dec. 2013, ISSN: 2234-0645. DOI: `10.5535/arm.2013.37.6.862`. [Online]. Available: `http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3895526&tool=pmcentrez&rendertype=abstract`.

[37] J. B. F. van Erp, F. L. Kooi, a. W. Bronkhorst, D. L. van Leeuwen, M. P. van Esch, and S. J. van Wijngaarden, “Multimodal Interfaces: A Framework Based on Modality Appropriateness,” *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, vol. 50, no. 16, pp. 1542–1546, Oct. 2006, ISSN: 1071-1813. DOI: `10.1177/154193120605001606`. [Online]. Available: `http://pro.sagepub.com/lookup/doi/10.1177/154193120605001606`.

[38] T. Takeya, “A Biofeedback Study of Postural Sway,” *Psychiatry and Clinical Neurosciences*, vol. 30, no. 4, pp. 495–504, Dec. 1976, ISSN: 1323-1316. [Online]. Available: `http://doi.wiley.com/10.1111/j.1440-1819.1976.tb02672.x`.

[39] R. A. Clark, R. McGough, and K. Paterson, “Reliability of an inexpensive and portable dynamic weight bearing asymmetry assessment system incorporating dual Nintendo Wii Balance Boards,” *Gait & Posture*, vol. 34, no. 2, pp. 288–291, 2011, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2011.04.010. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636211001378.

[40] L. Chiari, M. Dozza, A. Cappello, F. B. Horak, V. Macellari, and D. Giansanti, “Audio-biofeedback for balance improvement: an accelerometry-based system,” *IEEE Transactions on Biomedical Engineering*, vol. 52, no. 12, pp. 2108–2111, 2005, ISSN: 00189294. DOI: 10.1109/TBME.2005.857673. [Online]. Available: http://www.ncbi.nlm.nih.gov/entrez/query.fcgi?cmd=Retrieve&db=PubMed&dopt=Citation&list_uids=16366234.

[41] F. Avanzini, A. D. Götzen, S. Spagnol, and A. Rodá, “Integrating auditory feedback in motor rehabilitation systems,” *Proceedings of International Conference on Multimodal Interfaces for Skills Transfer (SKILLS09)*, 2009. [Online]. Available: http://www.dei.unipd.it/$%5Csim$spagnols/pubbl/SKILLS09.pdf.

[42] K. Vogt, D. Pirrò, I. Kobenz, R. Höldrich, and G. Eckel, “PhysioSonic-evaluated movement sonification as auditory feedback in physiotherapy,” in *Auditory Display - 6th International Symposium, CMMR/ICAD 2009, Copenhagen, Denmark, May 18-22, 2009. Revised Papers*, 2010, pp. 103–120. [Online]. Available: http://www.springerlink.com/index/r28777q20j73l7m1.pdf.

[43] G. Dubus and R. Bresin, “A systematic review of mapping strategies for the sonification of physical quantities.,” *PloS one*, vol. 8, no. 12, e82491, Jan. 2013, ISSN: 1932-6203. DOI: 10.1371/journal.pone.0082491. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=3866150&tool=pmcentrez&rendertype=abstract.

[44] T. Hermann, “Taxonomy and definitions for sonification and auditory display,” *Proceedings of the 14th International Conference on Auditory*

Display (ICAD 2008), B. Katz, Ed., pp. 1–8, 2008. [Online]. Available: http://www.icad.org/Proceedings/2008/Hermann2008.pdf.

[45] A. A. Gopalai and S. M. N. A. Senanayake, “Assistive Vibrotactile Biofeedback System for Postural Control on Perturbed Surface,” *Journal of Mechanics in Medicine and Biology*, vol. 13, no. 01, p. 1 350 006, Feb. 2013, ISSN: 0219-5194. DOI: 10.1142/S0219519413500061. [Online]. Available: http://www.worldscientific.com/doi/abs/10.1142/S0219519413500061.

[46] B.-C. Lee, “Design and Assessment of Vibrotactile Biofeedback and Instructional Systems for Balance Rehabilitation Applications,” Ph.D. Thesis. University of Michigan, 2012.

[47] C. Ho, H. Z. Tan, and C. Spence, “The differential effect of vibrotactile and auditory cues on visual spatial attention.,” *Ergonomics*, vol. 49, no. 7, pp. 724–38, Jun. 2006, ISSN: 0014-0139. DOI: 10.1080/00140130600589887. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16720531.

[48] B.-C. Lee, B. J. Martin, and K. H. Sienko, “Directional postural responses induced by vibrotactile stimulations applied to the torso.,” *Experimental brain research*, vol. 222, no. 4, pp. 471–82, Oct. 2012, ISSN: 1432-1106. DOI: 10.1007/s00221-012-3233-2. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/22968737.

[49] M. Azadi and L. Jones, “Evaluating vibrotactile dimensions for the design of tactons.,” *IEEE transactions on haptics*, vol. 7, no. 1, pp. 14–23, 2014. [Online]. Available: http://europepmc.org/abstract/MED/24845742.

[50] M. Dozza, C. I. Wall, R. J. Peterka, L. Chiari, and F. B. Horak, “Effects of practicing tandem gait with and without vibrotactile biofeedback in subjects with unilateral vestibular loss,” *Journal of Vestibular Research*, vol. 17, no. 4, pp. 195–204, 2007.

[51] L. a. Jones and N. B. Sarter, “Tactile Displays: Guidance for Their Design and Application,” *Human Factors: The Journal of the Human Factors and Ergonomics Society*, vol. 50, no. 1, pp. 90–111, Feb. 2008,

ISSN: 00187208. DOI: 10.1518/001872008X250638. [Online]. Available: http://hfs.sagepub.com/cgi/doi/10.1518/001872008X250638.

[52] S. K. Nagel, C. Carl, T. Kringe, R. Märtin, and P. König, "Beyond sensory substitution–learning the sixth sense.," *Journal of Neural Engineering*, vol. 2, no. 4, R13–26, Dec. 2005, ISSN: 1741-2560. DOI: 10.1088/1741-2560/2/4/R02. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16317228.

[53] Y. Visell, "Tactile sensory substitution: Models for enaction in HCI," *Interacting with Computers*, vol. 21, no. 1-2, pp. 38–53, Jan. 2009, ISSN: 09535438. DOI: 10.1016/j.intcom.2008.08.004. [Online]. Available: http://dl.acm.org/citation.cfm?id=1507761.1507842%20http://dx.doi.org/10.1016/j.intcom.2008.08.004.

[54] C. I. Wall and M. Weinberg, "Balance Prostheses for Postural Control," *Engineering in Medicine and Biology Magazine*, no. April, 2003. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=1195701.

[55] C. I. Wall, N. D. Lyford, K. H. Sienko, and M. D. Balkwill, "The design and development of a production prototype balance belt," *Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE*, vol. 2011, pp. 3524–8, Jan. 2011, ISSN: 1557-170X. DOI: 10.1109/IEMBS.2011.6090585. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/22255100.

[56] S. Kok-soon, L. Ming-Yih, L. Chih-feng, and T. Wen-Wei, "A new trunk sway assessment protocol using biofeedback inertial-based sensing modality for stroke patients," in *System Science and Engineering (ICSSE), 2011 International Conference on*, Jun. 2011, pp. 675–678, ISBN: 9781612844718. DOI: 10.1109/ICSSE.2011.5961988.

[57] J. Rosenthal, N. Edwards, D. Villanueva, S. Krishna, T. McDaniel, and S. Panchanathan, "Design, Implementation, and Case Study of a Pragmatic Vibrotactile Belt," *IEEE Transactions on Instrumentation and Measurement*, vol. 60, no. 1, pp. 114–125, Jan. 2011, ISSN: 0018-9456. DOI:

10.1109/TIM.2010.2065830. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5659690.

[58] Y. Leng, J. Xing, W. Zhao, M. Ji, and J. Xiao, "A noninvasive vibrotactile stimulator prosthesis for bilateral vestibular dysfunction," in *2013 1st International Symposium on Future Information and Communication Technologies for Ubiquitous HealthCare, Ubi-HealthTech 2013*, 2013, ISBN: 9781479907663. DOI: 10.1109/Ubi-HealthTech.2013.6708054.

[59] H.-Y. Yao and V. Hayward, "Design and analysis of a recoil-type vibrotactile transducer.," *The Journal of the Acoustical Society of America*, vol. 128, no. 2, pp. 619–27, Aug. 2010, ISSN: 1520-8524. DOI: 10.1121/1.3458852.

[60] S. Barrass and T. Barrass, "Embedding Sonifications in Things," *Proceedings of the 19th International Conference on Auditory Display (ICAD 2013)*, pp. 149–152, 2013.

[61] V. Hayward and K. E. Maclean, "Do It Yourself Haptics: Part I," *IEEE Robotics & Automation Magazine*, vol. 14, no. 4, pp. 88–104, Dec. 2007, ISSN: 1070-9932. DOI: 10.1109/M-RA.2007.907921. [Online]. Available: http://ieeexplore.ieee.org/xpl/freeabs_all.jsp?arnumber=4437756.

[62] J. van der Linden, E. Schoonderwaldt, J. Bird, J. V. D. Linden, W. Hall, and M. Keynes, "Good vibrations: Guiding body movements with vibrotactile feedback," in *Proceedings of the Third International Workshop on Physicality*, Sep. 2009, pp. 13–18. [Online]. Available: http://oro.open.ac.uk/19251/.

[63] J. McCartney, "SuperCollider, a New Real Time Synthesis Language," in *Proceedings of the 1996 International Computer Music Conference, ICMC 1996, Hong Kong, August 19-24, 1996*, 1996, pp. 257–258. [Online]. Available: http://hdl.handle.net/2027/spo.bbp2372.1996.078.

[64] J. R. Blum and J. R. Cooperstock, "Expressing Human State via Parameterized Haptic Feedback for Mobile Remote Implicit Communication," *Proceedings of the 7th Augmented Human International Conference 2016*

on - AH '16, no. Figure 2, pp. 1–2, 2016. DOI: 10.1145/2875194.2875225. [Online]. Available: http://dl.acm.org/citation.cfm?doid=2875194.2875225.

4 Modular Development Tools for Wearable Haptics (I 95%, S 75%)

(
words)

4.1 Approach

(ToDo: *next: fill in stuff from existing text, revise opener section*)

4.1.1 Lessons Learned from the Case Studies

4.1.2 Background: What happened in between?

Qwiic

4.1.3 Design of the modular toolkit

what constitutes the research contribution? reality-proven modular concept for in-situ wearable haptic sensor-actor systems -> identification of primitives, tools that fill the gaps, baseline experiments

Problem Definition and Intended Applications

Describe the system's requirements in terms of goals, non-goals and constraints; understanding an application domain, developing knowledge of hardware, software, human factors and algorithmic techniques in play is part of the contribution! Define the requirements of a "good" solution by communicating the key concerns of the problem space. Unexpected constraints and requirements reveal themselves only when real end-to-end systems are built!

provide a framework for assessing the quality of the system design decisions: are the design decisions made by the authors the reason why the system achieved the goals? match in evaluation

- rapid prototyping of belt, lra belt, gloves

context (to lead to more generalizeable knowledge)

- two main applications:
  - vibrotactile augmented feedback on posture in physiotherapy excercises: support rehabilitation outcomes, motivation, and adherence (primary user: physiotherapist, rehabilitation scientists)
  - vibrotactile brain-computer interfaces for minimally responsive patients: recognize most responsive episodes and provide a vibrotactile speller with

body-scheme training to maximize communication over the remaining channels (primary users: neuropsychologist, arbitrary minimally responsive patients)

- in both cases:
  - system needs to be adjustable/reconfigurable to the patient in question
  - choice of sensors depends on specific case/application
  - long-term studies are needed to validate laboratory success; hardware for this is not really available on the market
  - primary users are not computer engineers
  - systems need to be reliable enough to be used without supervision for long-term studies

Goals

- facilitate prototyping of sensor-actor wearables
- should be useable by technology interested, but not trained users
- leverage arduino and rapid prototyping ecosystem - open, modular, interoperable, ecosystem - use with multiple microcontroller architectures
- high fidelity haptics and audio on the body
- applicable to many applications, similar to brix2 but more integrateable for wearables
- support both low-end and high-end computational infrastructure
- interconnect: robustness, cost, assemble without special tools for arbitrary lengths
- low-latency embedded synthesis for closed-loop biofeedback systems
- optional connection to smart phones
- use cases

non-goals

- favour robust wire interconnect over wireless option for robust prototype systems

constraints

- reproduceable at reasonable cost for arbitrary researchers
- replacing / modifying cables
- integration into garmets, such as gloves, means compactness is key, in all axes

Key Insight

often central intellectual idea about the structure of a problem

Through our work on haptic wearables, here the belt and shoe described in the previous chapters, we have gathered knowledge on what is necessary for rapid development of wearable haptic prototypes.

it became apparent that for an effective development cycle the modularity of the system is key. Modularization enables reuse of parts of the system for different projects, and thus make recouperation of development and learning efforts more likely. As the individual components can be tested and modified in separation, repair and debugging are greatly facilitated. This is especially key for wearable development, where mechanical stress and the iterative, individualized design of an effective integration in to clothing are prone to break things. This process is necessarily a collaborative one, and a modular system allows parallel development work across multiple researchers, disciplines and institutes. A modular system also allows new extensions to be rapidly built, for example of freshly released technology or to adapt to the progressing research objectives.

Basing the sytem on an open hardware and software philosophy is contributing back to the community that enabled us to create our sytem and it means that we can discuss issues much more openly with upstream developers. It ensures the ability to maintain a growing ecosystem amongs multiple contributors with clear separation of what is considered an idea worthwhile of protection, e.g., through patenting, and what is not. The open source and open hardware movement has enabled accessible platforms such as Arduino without which development of embedded systems would still be limited to those wich profound expertise in electrical engineering and programming. As the development of effective wearables ties together expertise from many very different disciplines, it is paramount to lower the bar of entry and provide tools that can be taught outside of engineering and computer science schools. Using open interconnect standards, we can also integrate modules by others, further reducing development effort.

In this chapter, we will discuss three main contributions to this end. (ToDo: *this needs to be reworked*) We will discuss our initial extensions to an existing open hardware development platform, $BRIX_2$. This platform aimed at teaching and rapid prototyping, but the interconnect and housing proved too bulky for good integration into clothing. Thus, in the second half of the chapter, we will detail our new platform, based on the current state-of-the art technology with large-scale modularity integration into wearables as a key design objective.

focus on two types of actuators not yet easy to useable:

- LRAs for compactness
- Voice Coils / Lofelt T5 for strength, expressivity

modular system:

- interative development,
- extensible,
- repairable
- individual solutions - no user is the same

open hardware: embrace it

- hardware is hard, and this kind of research is just emerging - single research can not tackle the complexity of wearables
- openness allows continual improvement of basic primitives, IP is in the application
- interoperable, open standards create a marketplace that fosters innovation and competition: Smart Prototyping

Key design decisions (rationale, provide wisdom to future implementors)

"one common property of a preferred solution is that it is a simple one": eschew complexity that is not fundamental to meeting goals

key decisions (separate from implementation details, motivated by end-to-end view of the problem):

- no bluetooth for interconnect, no custom rf - latency/throughput are not good enough
- use existing interconnect system:
  - qwiic: market leader solution, but also most minimal connector (size) and pin count (no interrupts)
  - 4 pins are also great for two power rails, power and signal for actuators, uart
  - one cable for everything means that cables can be interchanged (for better or worse)
- parametric cad model for belt mounts

potential solution strategies / alternatives:

- use qwiic modules from sparkfun: too big, focus on being breakout boards
- modular prototyping from sparkfun, seeed: again, modules are huge and optimized to work as breakout boards as well - here we want highly integrated boards that can be integrated
- brix2 was modular, but has only 3 extension slots and a non-generic, high density connector that establishes mechanical stability through lego, this means it can not easily be integrated
- apdm is sensors only, no feedback

What were the alternatives considered at various points, and why were the choices made the way they were? Which side-streets did you not happen to explore? - ble nano: nrf5x is great for low-power systems, but shows how much footwork is needed to make development on a new ARM based architecture possible; maintenance is worth a lot, and simple programming over usb was deemed more important than smallest size (hence adafruit)

The driving motivation for the new design was to enable high-fidelity synthesis for the next generation of shoe and belt prototypes, while laying the foundation for a more modular architecture that could be extended for the specific applications.

In the LRA-based systems described in Section 3.6 and **??**, the synthesis and amplification are handled by a combined IC (TI DRV2605L). Here, we synthesize on the DSP extensions of the K20 processor, and the line-level signal needs to be amplified to drive the high-fidelity tactors. When presented with the choice of integrating the actuators on the synthesis board versus integration with the tactors, we chose the latter. D-class amplifier integration guidelines dictate short output wires to reduce EMI, as the output is a square wave, and we wanted to be able to use an arbitrary number of tactors in the future. Thus, we externalized the amplifier circuitry onto a separate board with small form-factor, described in Section **??**, along with the integration of the tactors in Section **??**.

All wearable systems need to be powered by battery, and we created a power board combining voltage regulator, power management and battery charging. Section 3.3.6 describes the module.

The interconnect of such a modular system is a critical part, as already discussed in Section B.3.3. The first version described in the belt chapter, 3.3.2, used custom cables, the second, described in Section 4.1.3, employs pre-made commercially available cables.

For the second generation of the system, we primarily aimed to improve the interconnect system and make the system even more modular and thus suitable for applications other than the shoe platform as well. We externalized the analog input for the FSRs onto a separate module with an ADC. Similarly, we placed the motion sensor on an extension, in the hopes of addressing the interference problems by physically separating the two parts of the board.

We based our interconnect on the recently released Sparkfun Qwiic system,[1] a four-wire I²C bus system based on JST SH (1 mm pitch) connectors. The individual leads are color coded (Black = Ground, Red = +3.3V Power, Blue = SDA / Serial Data, Yellow = SCL / Serial clock, and the connectors are polarized connectors to prevent mistakes. Pre-crimped cables for this system can be purchased through many distributors, and the extensions commonly two connectors on each extension to allow daisy chaining for rapid prototyping and reconfiguration with minimal cabling. This solves one of the biggest problems of modular prototyping, a compact connector and cabling solution that is widely available and standardized. Sparkfun pioneered the system with the express intent of becoming an open standard, and others have already followed suit in providing extensions for the system, which can readily be

[1] `https://sparkfun.com/qwiic`

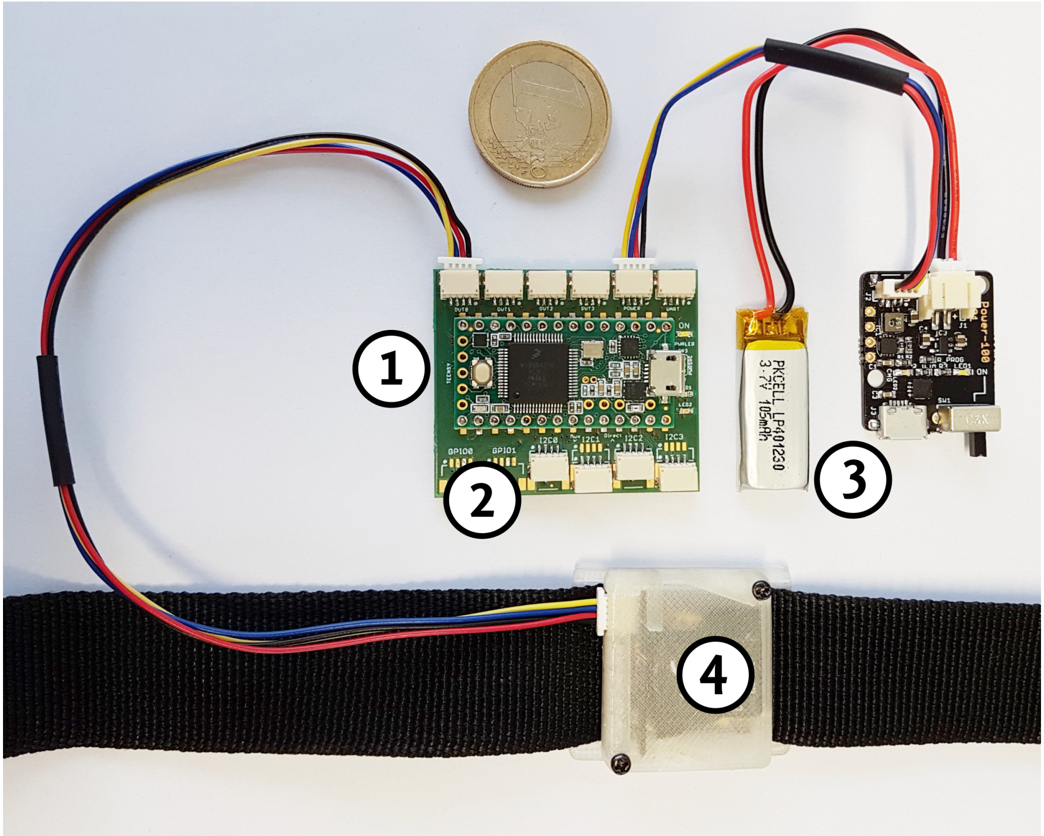


Figure 4.1: Second generation modular system example configuration, with one tactor and power supply. (1) Audio-400 base module, (2) Qwiic and general purpose input/ouput extension headers, (3) Power-100 power supply and lithium polymer battery, (4) Amp-200 amplifier module in Lofelt tactor in belt housing.

used with our base module. We expect this system to be one, if not the relevant standard for modular prototyping in the next years. There exist adapter cables to other modular prototyping systems, and existing modules can easily be connected through an adapter.

Big Todos

- lit review on wearable / vibrotactile prototyping (szehe thesis)
- lra haptuators were better than erms?

Ideas that have no home yet

- WebAudio may be good enough for future solutions (only 2 channels)

4.2 Module Overview

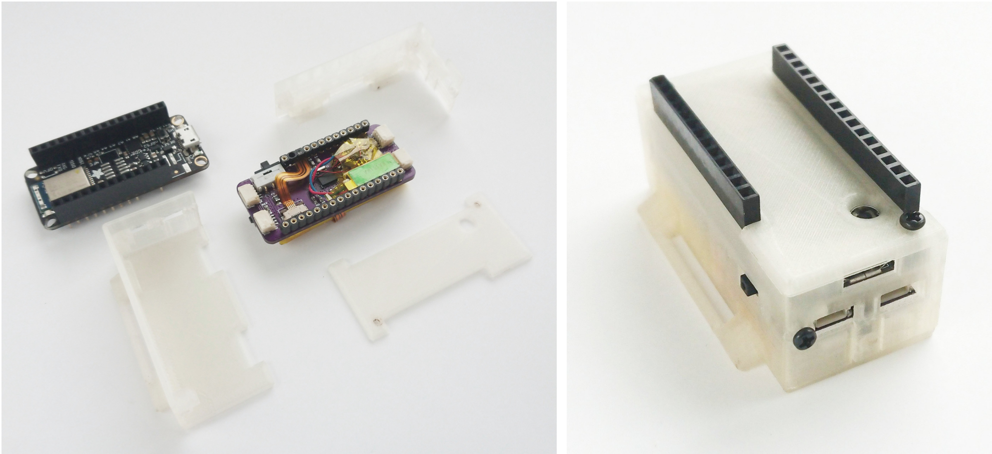
4.2.1 Base Controller

For untethered applications, a microcontroller controlling sensors, actuators, and communication needs to be placed on the body. Here, we present a flexible base controller that combines all ingredients needed for vibrotactile wearables. We base our design on two major advances of the open hardware prototyping community: the feather wing system, introduced by Adafruit[2], and the Qwiic system. These two open standards bridge between systems by these currently largest manufacturers of DIY electronics prototyping tools. The Feather system specifies[3] a rectangular footprint (0.9" x 2.0") for small microcontroller breakout boards that have a USB interface and include a LiPo battery power supply with charging circuit. Feather boards come with various microcontrollers, ranging from Atmel AVR 8-bit controllers to 32 bit Cortex M controllers, thus providing great flexibility for our base module. There support various wireless communication protocols depending on the microcontroller, e.g., Bluetooth Low Energy, WIFI, LoRa, XBee, and cellular networks 3G/4G.

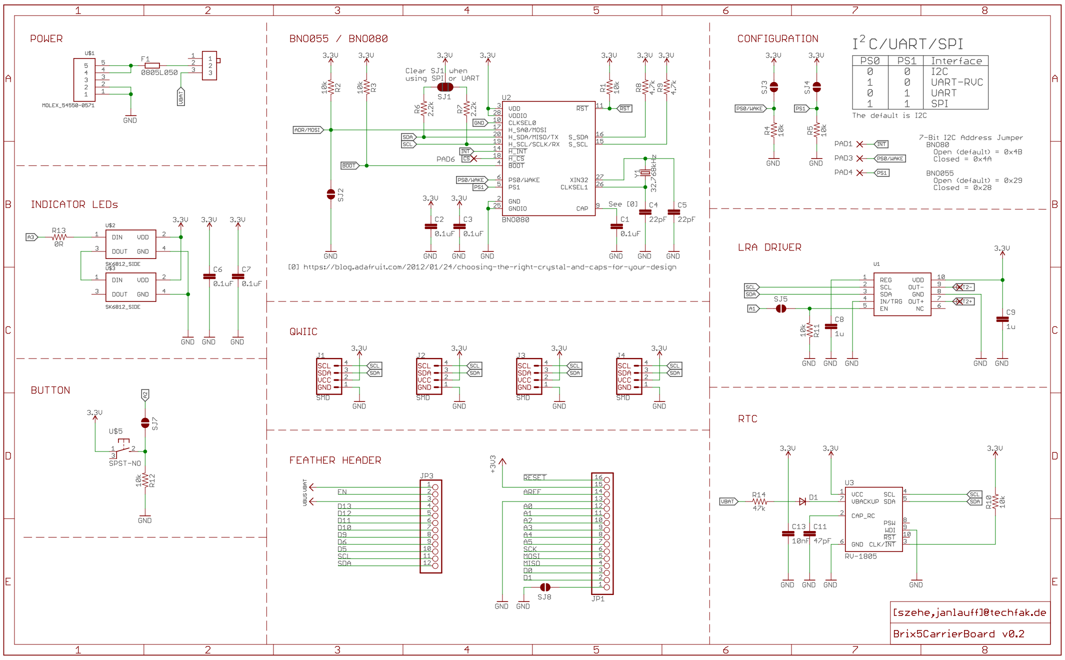
Feather boards can be extended with *Wings*, that stack on top or underneath. Our base module, shown in figure is a Feather with nRF52 or ESP32 microcontroller (others would work as well), housed together with a Feather Wing, that extends the basic breakouts with commonly required features required to build wearable applications:

- Four Qwiic ports for extension.
- A 450 mAh lithium polymer battery and a power switch
- A button, for user input
- A Bosch BNO055 motion/orientation sensor, see Section **??**
- A TI DRV2605 haptic driver (see Section 2.4.4), and a Jin Long Machiner xx LRA.

[2]`https://adafruit.com/feather`
[3]`https://learn.adafruit.com/adafruit-feather/feather-specification`



(a) nRF52 Feather with our extension wing, disassembled and in case



(b) Schematic

Figure 4.2: nRF52 Feather Based Main Module

- Two addressable RGB indicator LEDs, for example to provide detailed information on system status
- An RV1805 low power, high preceision real-time clock for real time stamps on sensor data

We were inspired by the wealth of wearables, that include BLE communication, motion sensors, and often an ERM tactor, but lack customization options. To our knowledge and at the time of writing, no system exist that provide a modular basis for wearable prototyping. It is also an evolution out of the nRF51-based, stackable prototypes that were described in Section 3.6. The base module is highly integrated and xx in size, and weights xx. (ToDo: *dimensions*) Our wing preserves the ability to extend the Feather with other wings, and together with Qwiic-compatible ones, dozens of extensions exist. In the following, we describe some that we added.

4.2.2 Teensy-based Audio-400 Base Module

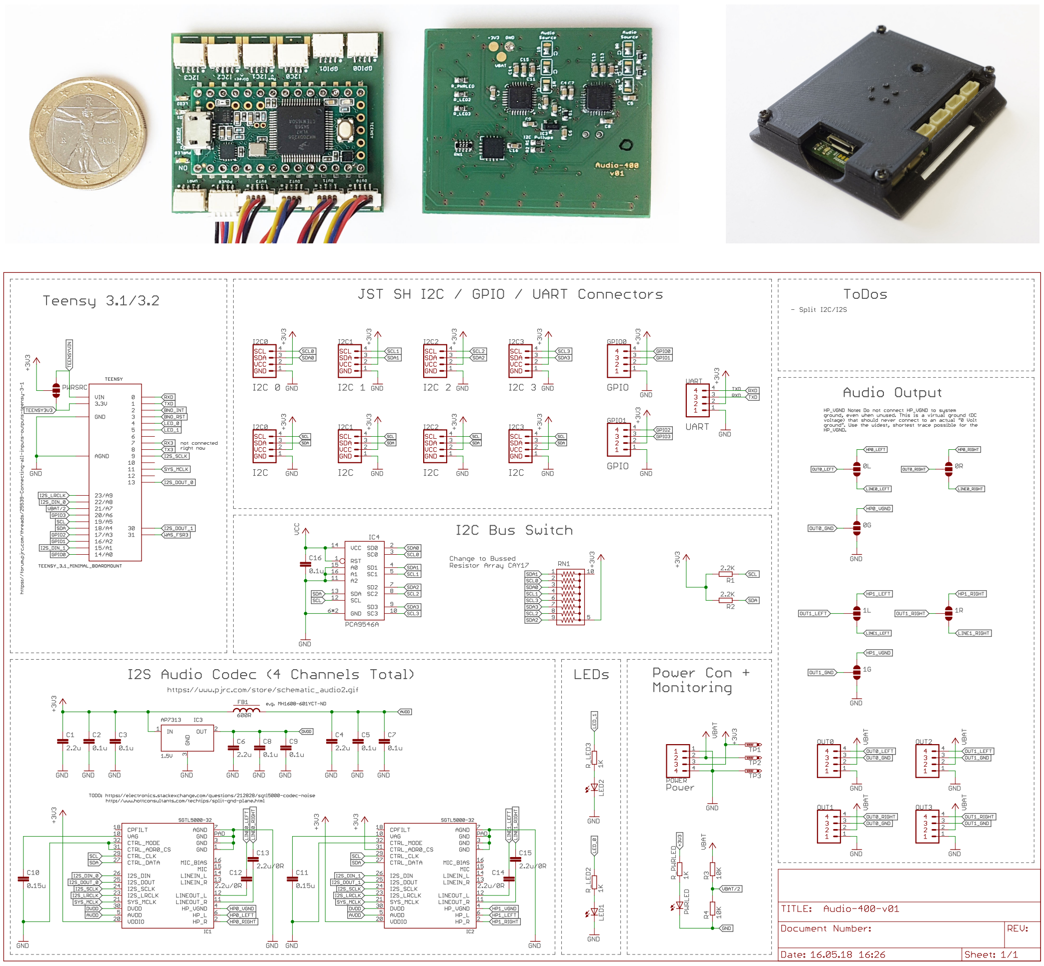


Figure 4.3: Audio-400 base module: PCB top/bottom view, in belt housing, schematic

The Audio-400 base module is shown in Figure 4.1 and 4.3. This module is directly based on the previous version (TT-300) with a number of significant improvements. We decided to directly solder the Teensy on the base module, as to reduce vertical height. The two pins on the back side of the Teensy, required for four-channel synthesis, are soldered through holes in the base module. The module's height was reduced from 19 mm to 6 mm (excluding the DF-17 $BRIX_2$ connector on the first generation).

We used the Qwiic system's JST SH connectors for all connections on the board to leverage the availability of a compact interconnect fully. Two general-purpose input/output (GPIO) ports break out two digital / analog pins from the Teensy plus +3.3V/ground, each. The UART serial port, for example to connect a bluetooth module, the power input, and the audio outputs all use the SH connector as well.

In order to be able to connect multiple sensors of one type, that may only have one or two I^2C bus addresses, we included a four-channel I^2C bus multiplexer chip on the base module (NXP PCA9546). Qwiic extension can either be connected directly to the Teensy, or through the multiplexer; the SH headers have two overlapping positions that can be populated depending on the application.

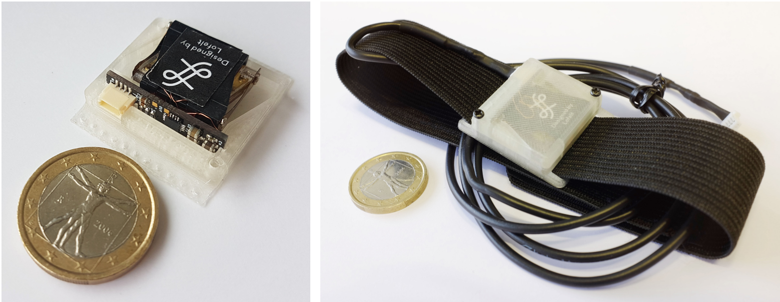
4.2.3 Amplifier Module Amp-200 and improved housing

As the Lofelt actuators, described in Section 3.3.3, are larger and can handle more input power, we needed to increase the power of the amplifier module. We chose the Maxim MAX98307 3.3W class DG amplifier,[4] which features an integrated charge pump that allows it to run directly from the unregulated battery voltage and also enables high efficiency over a wide output power range. The chip also includes active EMI limitting circuitry.

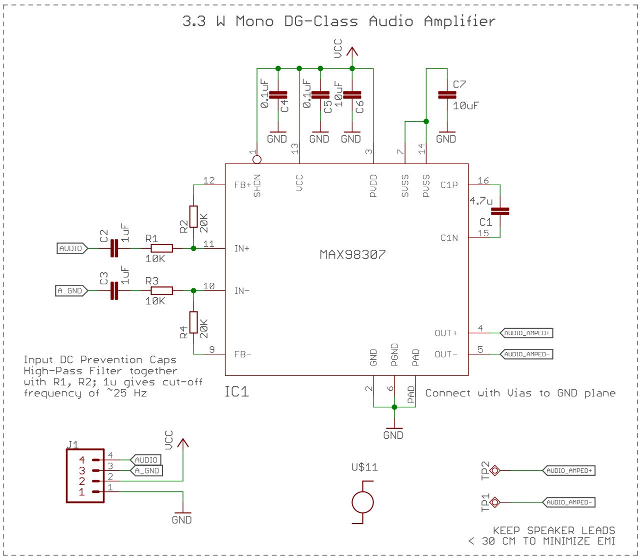
Instead of being stacked over the actuator in the tactor module, we decided to place it vertically on the same plane to reduce tactor height and improve cable routing. The highly integrated module is only 25×6 mm² in size and manufactured using 1 mm thick PCB material, the combination of Lofelt actuator and amplifier board can be housed in a very compact enclosure. Figure 4.4 shows integration in a two enclosures, one for sewing onto garments, and the schematic.

Power Consumption: 4x White Noise @ 217 mA

[4] https://www.maximintegrated.com/en/products/analog/audio/MAX98307.html



(a) Amplifier module Amp-200 in sewable enclosure with Lofelt actuator



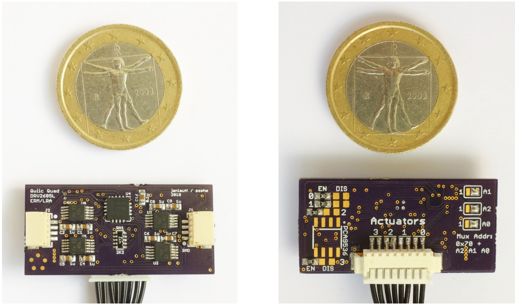
(b) Amp-200 Schematic

Figure 4.4: Audio Connector and Distributor Board

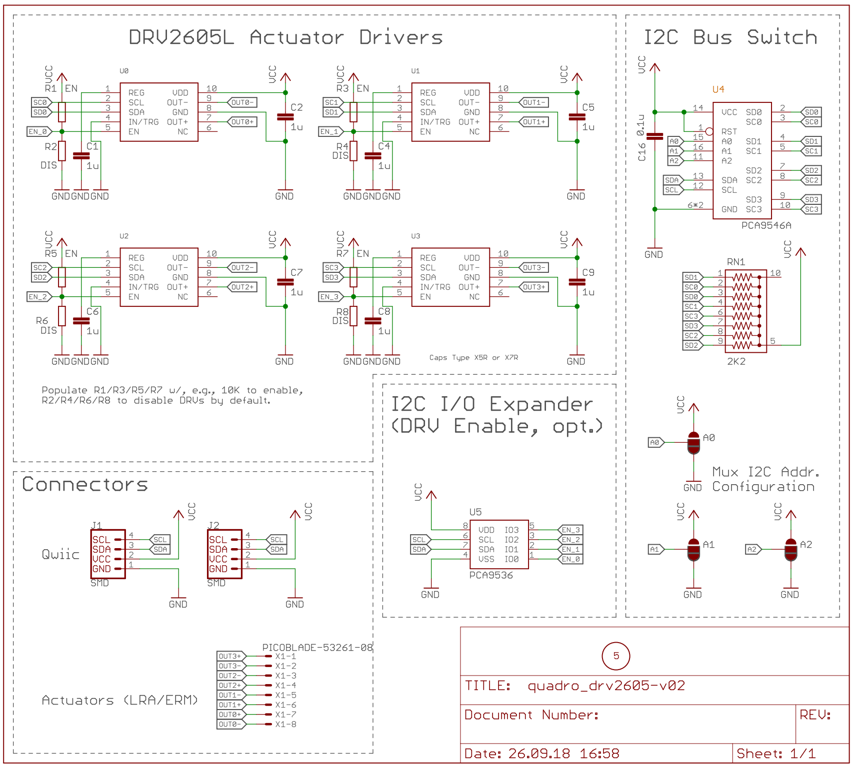
4.2.4 Quad-Channel ERM/LRA Driver

We developed a Qwiic-compatible quad-channel LRA driver module, based on the tacton shoe design described in Section B.4.3 and the BLE Nano extension discussed in Section ??. At its core are four Texas Instruments DRV2605L tactor drivers that support both ERMs and LRAs and is able to drive them with overdrive and braking for maximal actuator expressiveness. LRAs are driven with an AC signal that needs to match the resonance frequency of the actuator for maximum efficiency; the DRV2605 tracks this frequency automatically and delivers waveforms appropriate to each type of actuator. It includes a built-in library of 123 haptic effects by Immersion and can also drive the actuator based on a time-varying I^2C intensity, PWM or analog input; here, we do not break out these

An I2C bus switch / multiplexer is required to use more than one chip in a design, such as the Texas Instruments TCA9685 used here to arbitrate between the system I^2C bus and the individual drivers. The multiplexers I^2C bus address can be configured to be one of eight choices, for a total of 32 LRA tactors that could be realized on one bus. An optional TI PCA9536 4-Channel IO expander`http://www.ti.com/lit/ds/symlink/pca9536.pdf` provides the option to disable individual DRV2605L chips in order to save power when they are not used. The custom PCB is shown in Fig. 4.5 4.5a, the schematic in Fig. 4.5 4.5b. To our knowledge, there is no multi-channel tactor driver board available for Qwiic on the market. It enables rapid prototyping of multi-actuator wearables, as demonstrated with the NeuroComm-Trainer glove described in Section D.1.3 and the LRA haptic belt described in Section **??**. We modified the Adafruit DRV2605 Arduino library to provide support for multiple actuators (ToDo: *link to repo*).



(a) PCB, top + bottom



(b) Schematic

Figure 4.5: Qwiic Quad LRA/ERM driver module

4.2.5 Motion Sensor Modules

(ToDo: *explain requirements, attempts etc for this*) (ToDo: *make this about bno055 as well*) (ToDo: *move nxp to examples for custom extensions*) A key component of wearable computing are motion / orientation sensors. With the advent of smartphones, these have become increasingly integrated and affordable; and subsequently very attractive for implementation of all kinds of wearable applications. Our sensor of choice is the Bosch Sensortec BNO055/BNO080 motion sensor. It incorporates 3-axis accelerometer, gyrometer and magnetometer sensors with an integrated sensor fusion for a stable orientation reading. In comparison to other chips of this kind, namely the Invensense MPU9250 $BRIX_2$ was based on, the output is surprisingly drift-free as it can compensate for chip temperature (ToDo: *verify*). The BNO080 is mostly pin-compatible with the BNO055 but has improved orientation computation sensor-fusion algorithms operating at 400 instead of Hertz, designed with virtual reality applications in mind.

(ToDo: *somewhere, we should explain all the different things we tried that failed, including the nxp fuckage*) Our PCB design can be used for both chips, the schematic in shown in Figure 4.6 (d). We developed three different variants of the same design. The most commonly used one, shown in Fig. 4.6 (a)/(b) exposes most configuration options and can be daisy-chained.[5] For integration in very confined spaces, such as the tips of gloves, we shrunk the design even further, dubbed MicroBNO, shown in Fig. 4.6 (c). A third variant forms the basis for the Adafruit Feather Qwiic wing, described in (ToDo: *ref section*).

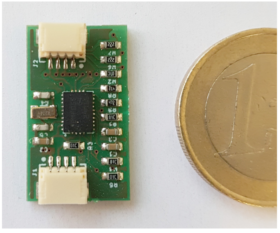
4.2.6 Analog Input Module (Analog-100)

This module contains an I²C analog-digital converter and the opamps for an current-to-voltage converter to condition the signal of force-sensing resistors. Both the 16-bit Analog Devices ADS1115 and 12-bit ADS1015 (higher sample rate) can be populated. Teensy has integrated ADCs, and we created this extension to incorporate the signal condition into the system off the main unit to reduce the amount of input connectors and components on the Analog-400 module. This also brings the ADC closer to the sensors that are being read. This module is a miniaturized version of an Adafruit product[6] that we extended with the opamp circuitry and solderjumpers to enable all possible I²C bus addresses. Solderjumpers also allow to bypass the opamps.

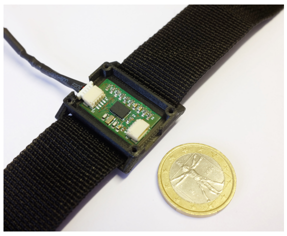
4.2.7 Other Microcontroller Options

- ESP32 via Adafruit, Zio
- Raspberry Pi for Programming with Python

[5] Bus communication protocol selection, bus pullup resistor enable, chip address selection
[6] `https://www.adafruit.com/product/1085`

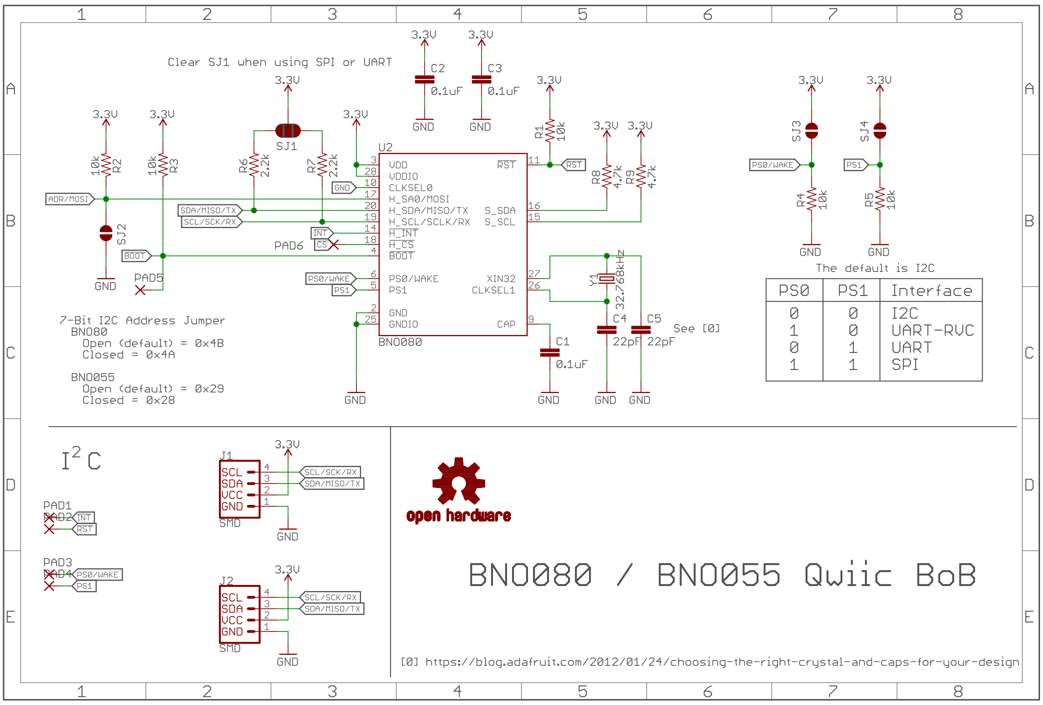


(a) Basic Extension



(b) Strap Mount

(c) Micro BNO



(d) General schematic

Figure 4.6: Bosch BNO055/BNO080 motion sensor extensions

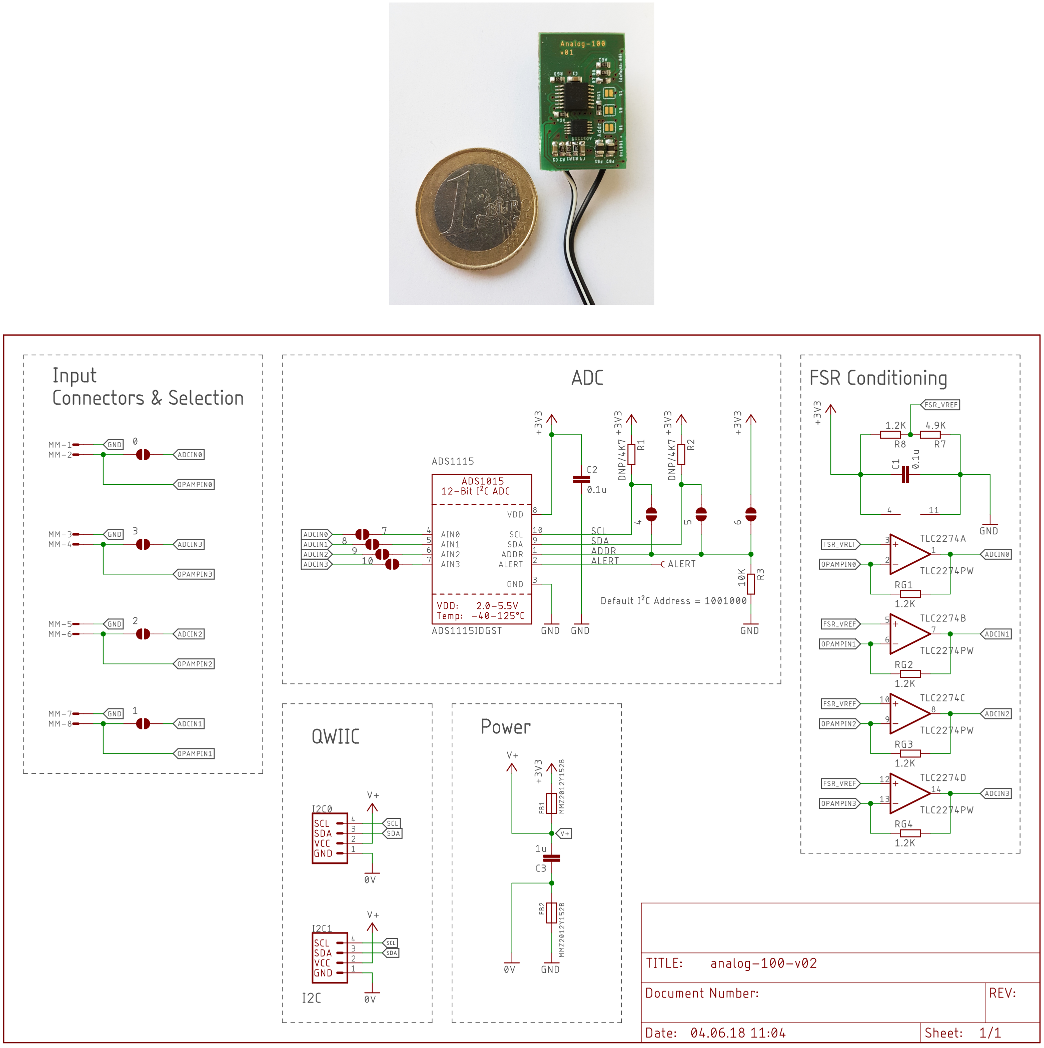


Figure 4.7: Analog input module (Analog-100)

4.3 Wireless Communication and Data Logging

The communication between host computer and haptic wearables is through a Bluetooth Low Energy (BLE) link, through the commonly used Nordic UART Service (NUS). This proprietary BLE profile has become the defacto standard for mimicking the RFCOMM GATT profile of previous Bluetooth versions. Here, we are using Nordic NRF5x bluetooth chips on the wearable side, and both the Sandeep Mistry Arduino Core + BLEPeripheral library and the Adafruit fork implement the NUS profile. We chose to encode our communication through this simulated UART scheme for three reasons. First, implementation is greatly facilitated as no understanding of creating a proper BLE profile is needed, and it already comes tested and with support from upstream developers. Second, it allows us to operate the commands interactively by hand without needing to implement specific tooling. Third, it works just the same over a wired (USB) serial link, or, if BLE proves to be not a suitable wireless standard, over another packet radio link standard.

[Host Computer] <=> [nRF5x] (<=> [Teensy])

In cases where we use another microcontroller, such as the Teensy employed to synthesize stimuli for the LoSound voice coil actuators (the successor of the Basslet in our experiment), we can just use a NUS serial bridge to integrate it to our system as well. Dedicated modules for this purpose exist from Adafruit (Bluefruit LE), here, we chose to use Adafruit Feathers, in case we need additional I/O functionality. Previously, we employed RedBearLab BLE Nanos. In all cases, a simple program forwards each character received between hardware UART and NUS, bidirectionally. According to Adafruit, communication speeds over 9600 baud may lead to packet loss, previously, we saw little problems with a similar implementation at 38400 baud.

Implementation on the BLE central side, e.g., host computer or smartphone, then is with the commonly integrated Bluetooth functionality. We developed an implementation based on the Web Bluetooth standard, which runs (currently) on Chrome/Chromium, both on desktop computers and mobile devices, without requiring additional software. This replaces an initial implementation with Evothings, a Javascript binding to native Bluetooth functionality based on Phonegap. For non-interactive use, we based our demo implementation on the Adafruit Python library.

(ToDo: *no longer accurate*) In the NeuroCommTrainer (NCT) project, we have three devices communicating through BLE with a host computer, running the latter Python-based implementation:

- Two gloves, with 3 Linear Resonant Actuators (LRAs) for actuation and one Bosch BNO055 9 DoF MARG motion sensors each. - One four-channel LoSound voice-coil system, with one BNO055 per output channel.

Thus, we have three in-/output streams to deal with, possibly running in a separate instance of the python program (that has yet to be finally implemented). The embedded controllers parse input through the CmdArduino library, which means that commands are to be terminated with a carriage return character (˚). The command 'h' outputs an overview of all available options. For sensor readout, there are two commands, that would need to be triggered before the stimuli presentation, and again after, to conserve power by disabling sensor data streaming:

- 'B' toggles, 'B 1' enables, 'B 0' disables a HEX-encoded stream of sensor values. The format is still not completely defined, but here is an example: TTTTAAAABBBBCCCCDDDDCCCC, where TTTT is a time stamp and AAAA is a 16-bit HEX-encoded sensor value (limit of 20 bytes per BLE message)

TODO: Hier wird klar, das wir noch Arbeit zu tun haben: Der Timestamp ist noch nicht definiert (mit datum? unix millisekunden vll? Und was ist mit 4 sensoren am Teensy (gut das wir da die Muxe drauf haben!)? - 'A' toggles an ASCII encoded stream of sensor values (human readable, for debugging purposes)

For generating stimuli, we have to distinguish between the LRA and the voice-coil-based system. For the LRA systems, we can either choose effects from the integrated library (see Texas Instruments DRV2605L datasheet), or define arbitrary effects on any of the four output channels, A/B/C/D in the following (only three are planned to be used for the current demonstrator). As an example, 'p 1a' would present a single buzz on actuator A, 'p 3bcd' would play a sequence of tripple buzzes on actuators B, C and D. We still need to define the exact stimuli (e.g., length and intensity 0..255) for the demonstrator, to then associate them with new commands in similar vein. To control the voice coil actuators, a similar pattern will need to be defined for each stimuli that is to be presented. TODO: wir muessen noch die Stimuli von Laura definieren/rausfinden/implementieren.

The Teensy module can also report on battery charge (TODO: implementieren und rausfinden ob Feather das vll auch messen kann).

4.4 Outlook

4.4.1 Open Hardware

Release schematics, possibly get Adafruit to make them stuffs for us..

Power

Being modular also means the ability to replace the current module with one that has more power or supports charging battery chemistries different from lithium ion/polymer. The Microchip MCP1631[7] would be an option as a programmable charger supporting 1-2 Lithium and 1-4 Nickel Cadmium (NiCd/ Nickel Metal-Hydrid (NiMH) batteries in series. Make batteries possible to rip off the body if they catch fire.

Auto-Calibrated Actuator

This invention is comprised of a voice-coil type actuator in combination with a driver and synthesis board. This enables a bus- topology for connecting multi-channel haptic rendering devices. An integrated accelerometer enables the driving circuit to adapt amplification gain to an estimated coupling of the actuator to the user.

A. We are currently investigating wearable haptic systems. We collaborated with the company InterDigital on an ENGAGE in the domain of foot-based haptics. From previous researches, we determined that the existing haptic actuators are either not strong enough and have limited expressivity (Linear Resonant Actuators) or are difficult to integrate into wearables due to their need for signal synthesis and amplification for each channel. Previous research in collaboration with Jeff Blum regarding actuator coupling prediction was detailed in ROI 17008 and 15109.

B. We built our voice coil actuator using three basic materials: inductor coil former, magnet wire, and neodymium magnet. (See “Haptic Actuator Design” folder) A round amplifier PCB, which fits around the actuator, has been designed. The accelerometer and the microcontroller have not been integrated yet. (See “Audio Amplifier PCB” folder). The basis of the electronics comes from previous experience in the development of tiny electronics circuitry in the form of extensions for the BRIX development system. The round PCB insert also serves as a vibration baffle, improving actuator efficacy, and mechanical retainer increasing coupling surface. C. There are three main markets to which this invention applies. It applies in the research industry mainly for perception study, human to computer interaction, and haptic feedback. Another market share will be in the rehabilitation industry. This invention can be use to enhance the sense of touch for people who have disability which affect this sense. It can also be used in other rehabilitation task such as balance task for relearning to walk after an accident. A mass-market opportunity will come from the gaming industry. This invention can be used in many ways for virtual or augmented reality. D. Predicting coupling of actuators with humans is

[7] https://www.microchip.com/wwwproducts/en/en531783

an intrinsic and critical problem in wearable haptics, especially as the user moved and it is likely to change over time. This is why this invention propose a mechanism for compensating for the coupling discrepancies. Combining the circuitry required to synthesize and amplify appropriate signals with the actuator itself reduces the number of wires to a constant (bus signal lines, power), whereas before individual lines where necessary and that in turn might cause routing problems or electromagnetic interference. This, along with the low profile of the technology, enables us to engineer the next generation in wearable augmented and virtual reality, which was so far limited by the lack of bandwidth and strength of LRAs and the size of existing driver solutions for voice coils.

What is novel? The use of haptic illusion(s) to represent information in 3 dimensions around and or through the body (e.g., wrist, ankle, foot, etc.) using a small number of discrete vibration actuators. What are the uses Improved spatial resolution of haptic feedback, more communicative circular information representation, (applications)? simulation of complex haptic experiences requiring around or through-body actuation, How is it non- The illusions proposed are usually used in a 2D matrix representation and to date, we are not aware of obvious? any instances of circular (around the body) or through the body haptic systems that are employing the tactile illusions to increase their spatial resolution. 11. Publication/Disclosure: Have you published/disclosed any aspect of the invention? Are you planning to publish?

4.5 Publications and Disclosures

ROI Mcgill 17008: Anlauff, Blum, Cooperstock: Inferring haptic actuator coupling from actuator current draw.

ROI McGill 17035: Dubé, Anlauff, Cooperstock: Auto-calibrated Haptic Actuator with Integrated Signal Amplification

4.6 Acknowledgements

Sebastian Zehe and CITEC provided help with the assembly of the surface mounted boards presented in this chapter. He also contributed to several PCB designs, specifically routing ($BRIX_2$ audio v2, Audio-100 v01, Power-100 v01), the DB-100, and the design of the actuator enclosure. Bei Liu contributed to design and routing of the TT-300 and BLE Nano based boards.

5 Discussion

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words)

5.1 Outlook

5.1.1 Pattern Design and Authoring

The issue of signal generation relates to the higher-level requirement of suitable authoring tools for haptic effects. Tools for single [1] and multi-actuator [2] arrangements have been proposed by researchers and industry. We foresee a need for specialized authoring tools addressing the needs of the particular applications, for example taking into account the user's locomotion state.

5.1.2 Other applications

Exercise Feedback `https://ieeexplore.ieee.org/document/8329649` Back pain `https://tutcris.tut.fi/portal/files/16507236/A_Novel_Technique_for_Analysis_of_Postural_Information_with_Wearable_Devices_post_print.pdf` Gait training `https://jneuroengrehab.biomedcentral.com/articles/10.1186/s12984-017-0313-3`

Motivation: Improved Alarm Systems

[3], [4]

Our long-term vision is to suspend the rehabilitation support into the background, removing the need for conscious interaction.

Smart Textiles

Recently, textiles with integrated sensing abilities [5] have been announced, such as the pressure sensing Sensoria sock[1]. Similarly, Preece et al. introduced an instrumented sock for gait analysis b0ased on woven strain gauges [6]. Stoppa and Chiolerio reviewed the state of the art of such smart textiles [7] and we believe that such sensors will become more common in ubiquitous rehabilitation applications in the near future.

[1] `http://store.sensoriafitness.com/products/`

References for Chapter 5

[1] O. S. Schneider and K. E. MacLean, "Improvising design with a Haptic Instrument," in *Haptics Symposium (HAPTICS), 2014 IEEE*, Feb. 2014, pp. 327–332. DOI: 10.1109/HAPTICS.2014.6775476.

[2] S. Panëels, M. Anastassova, and L. Brunet, "TactiPEd: Easy prototyping of tactile patterns," in *Lecture Notes in Computer Science (including subseries Lecture Notes in Artificial Intelligence and Lecture Notes in Bioinformatics)*, 2013, ISBN: 9783642404795. DOI: 10.1007/978-3-642-40480-1_15.

[3] V. Cobus, B. Ehrhardt, W. Heuten, S. Boll, and W. Heuten, "Vibrotactile Alarm Display for Critical Care," in *Proceedings of the 7th ACM International Symposium on Pervasive Displays*, ser. PerDis '18, New York, NY, USA: ACM, 2018, 11:1–11:7, ISBN: 978-1-4503-5765-4. DOI: 10.1145/3205873.3205886. [Online]. Available: http://doi.acm.org/10.1145/3205873.3205886.

[4] V. Cobus, S. Boll, and W. Heuten, "Requirements for a Wearable Alarm Distribution System in Intensive Care Units," in *Zukunft der Pflege - Innovative Technologien für die Pflege*, Oldenburg, Germany, 2018, pp. 185–189.

[5] R. Paradiso, G. Loriga, and N. Taccini, "A wearable health care system based on knitted integrated sensors.," *IEEE transactions on information technology in biomedicine : a publication of the IEEE Engineering in Medicine and Biology Society*, vol. 9, no. 3, pp. 337–44, Sep. 2005, ISSN: 1089-7771. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16167687.

[6] S. J. Preece, L. P. J. Kenney, M. J. Major, T. Dias, E. Lay, and B. T. Fernandes, "Automatic identification of gait events using an instrumented sock.," *Journal of neuroengineering and rehabilitation*, vol. 8, p. 32, Jan. 2011, ISSN: 1743-0003. DOI: 10.1186/1743-0003-8-32. [Online]. Available: http://www.springerlink.com/content/d1412g20p8r22700/.

[7] M. Stoppa and A. Chiolerio, "Wearable electronics and smart textiles: a critical review.," *Sensors (Basel, Switzerland)*, vol. 14, no. 7, pp. 11 957–92, Jan. 2014, ISSN: 1424-8220. DOI: 10.3390/s140711957. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/25004153.

6 Conclusion

(
words)

Three Loops

(jan: *move this to the beginning of the balance chapter; or even to the conclusion?*) We can consider the problem of supporting in-home balance rehabilitation problems as a combination of three loops, pictured in Figure 6.1. Maintaining upright equilibrium, or balance, has been defined as the ability to maintain the body's centre of mass within manageable limits of the base of support, as in a standing or sitting posture, or control its transition to a new base of support, as in walking or moving [1]. Balance is considered to be a *multisensory integration* task, a complex closed-loop control system [2] represented by the innermost loop of Figure 6.1 that links perception and action. The central processes perceive individual visual, vestibular, tactile and proprioceptive sensations and generates task-specific and goal-directed movements to adapt to the environment [3].

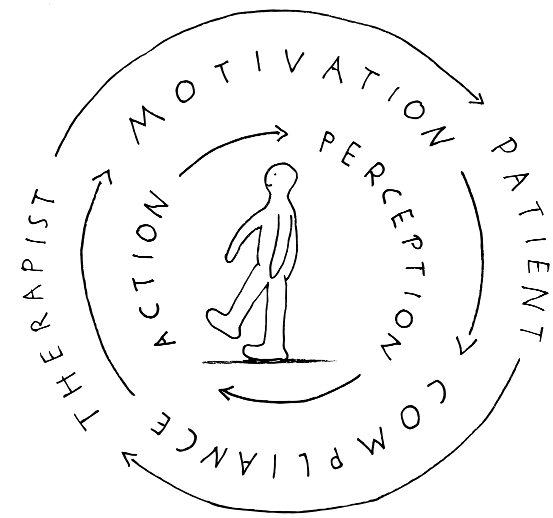


Figure 6.1: The three loops we propose to address in this thesis.

Traumatic brain injury, such as stroke, and losses of sensory function, such as aging and vestibular loss, can disrupt this ability. Physical and occupational therapists usually treat patients by developing a set of exercises aimed at helping the patient with everyday activities. An assessment of the patient's specific impairments of the sensory, motor, cognitive, musculoskeletal and other systems can guide the therapist to prescribe exercises, ideally along a progression that matches the rehabiliation ability of the individual(Add Citation: *Sienko's progression paper*).

Returning home, patients (however?) are mostly on their own. Staying motivated and doing the right exercises, and doing them right, in other words, staying *compliant* to the prescribed exercises, consequently has long been identified as a critical issue in rehabilitation [4]. This is especially crucial in the case of traumatic brain injuries, such as stroke, where interventions should be early, intensive and repetitive to be most effective(Add Citation: ?). The process might be complicated further by coinciding cognitive deficits or symptoms such as depression [5]. Individuals also may have a hard time perceiving their rehabilitation progress without external reference, as progress can be slow. Consequently, supporting the patient in a way that sustains both compliance and motivation is crucial for in-home rehabilitation systems to be effective. The interaction between motivation and compliance to the prescribed interventions is represented by the middle loop of Figure 6.1.

Finally, the therapist's ability to provide guidance, assess short- and long-term changes of the patient's capabilities [6], and accordingly adjust the prescribed exercises is currently limited to times of interaction during therapy sessions. This may in turn affect motivation and compliance negatively. The communication loop of patient-therapist interaction is represented by the outermost loop in Figure 6.1.

We summarize our goals with regard to to closing the three loops.

Augmented Feedback

For the innermost, sensorimotor integration loop, we propose *integration of multiple sensing and feedback modalities*, and the ability to *individualize* a *modular system* to adapt it to patients needs. As part of this system, we developed *wearables* combining motion sensing and *high-fidelity vibrotactile actuation* for the waist and the foot.

Increasing Motivation & Ensuring Compliance

For the middle, interaction of motivation and compliance, loop we propose a *balance game* that yields a *performance score*(jan: *can we claim this?*). *Serious games*, or *exergames*, in which the user controls video games by performing exercises can be highly engaging [7] while at the same time providing feedback on the execution quality of the movement to the user, and have been found to benefit motivation [8]. The process of applying game thinking and mechanics in non-game environments is called *gamification*. As initial application of our wearable building blocks system, we plan to develop a balance game that yields a *metric* of the user's day-to-day balance performance. We hope to make the game independent of a graphical display for the actual game mechanics, instead relying as much as possible on the haptic interaction with the user in combination with a smart watch for textual information.

Improving Patient & Therapist Communication

To improve the communication between therapist and patient, we will track the metric over time in order to quantify the day-to-day balance performance of the user along with their *annotations* (jan: *this does not really exist*). Storing the metric over time and providing access and analytics, both to patient and therapist, can provide valuable insight on the user's rehabilitation progress. The idea that "data creation, information generation, meaning-making, and action-taking" [9] can lead to lasting positive impact on society is described by the notion of a *quantified self*, and has recently been proposed for application in rehabilitation science [10]. The knowledge about the tracking of a performance metric could also result in a sense of accountability which may be beneficial to a patient's motivation. We plan to allow the patients to record their own annotations with regard to this metric, such as how they felt, what external circumstances influenced their balance that day, and their pain levels. Such a rich dataset could help guide the therapist in adjusting exercises and deciding when to empower the patient [11].

All three loops need to be addressed to bring effective biofeedback rehabilitation aids to the homes of patients. Most augmented feedback research to date is laboratory or clinic-based and the involved instrumentation is often expensive and complex. Video game based systems based on entertainment sensing technology such as the Kinect and Wii Balance board are showing promise to address these issues, but still require the user to be in the vicinity of a stationary system. In contrast, we propose to work towards *wearable*, self-contained systems that can be used in any environment, and can be used to support dynamic balance tasks such as walking. (ToDo: *where does this go: With a few notable exceptions [12], [13], studies were conducted under lab conditions.*) There has been some progress towards wearable systems in the recent years, but most studies were short in time and investigations on long-term effects are rare, as a recent review of feedback systems for gait showed [10]. This leaves efficacy and long-term effects largely unevaluated. Reliable and long-lasting self-contained systems are needed and to make these work outside of the clinic, and they have to be designed with the user in mind and evaluated in the real world. Others have established necessary criteria for the design of such wearables [14], [15]. One major limitation of existing wearable balance training systems is that they only work with a single type and unit of sensor. Our hope is that by addressing the problem of in-home balance rehabilitation on these three levels will enable progression towards such a ubiquitous, temporary-use system(jan: *was not introduced*), while providing distinct research contributions in themselves(jan: *this is thin ice*).

References for Chapter 6

[1] T. E. Howe, L. Rochester, A. Jackson, P. M. H. Banks, and V. A. Blair, "Exercise for improving balance in older people," *The Cochrane database of systematic reviews*, no. 4, p. CD004963, 2007, ISSN: 1469-493X. DOI: 10.1002/14651858.cd004963.pub2. [Online]. Available: http://dx.doi.org/10.1002/14651858.CD004963.pub2.

[2] M. E. Tyler, Y. Danilov, and P. Bach-Y-Rita, "Closing an open-loop control system: vestibular substitution through the tongue," *Journal of integrative neuroscience*, vol. 2, no. 2, pp. 159–164, 2003. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/15011268.

[3] N. Bugnariu and J. Fung, "Aging and selective sensorimotor strategies in the regulation of upright balance," *Journal of neuroengineering and rehabilitation*, vol. 4, p. 19, Jan. 2007, ISSN: 1743-0003. DOI: 10.1186/1743-0003-4-19. [Online]. Available: http://www.pubmedcentral.nih.gov/articlerender.fcgi?artid=1910603&tool=pmcentrez&rendertype=abstract.

[4] K. Andrews and J. Stewart, "Stroke recovery: He can but does he?" *Rheumatology and rehabilitation*, vol. 18, no. 1, pp. 43–8, Feb. 1979, ISSN: 0300-3396. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/424667.

[5] A. J. Carson, S. MacHale, K. Allen, S. M. Lawrie, M. Dennis, A. House, and M. Sharpe, "Depression after stroke and lesion location: a systematic review.," *Lancet*, vol. 356, no. 9224, pp. 122–6, Jul. 2000, ISSN: 0140-6736. DOI: 10.1016/S0140-6736(00)02448-X. [Online]. Available: http://dx.doi.org/10.1016/S0140-6736(00)02448-X.

[6] B.-C. Lee, "Design and Assessment of Vibrotactile Biofeedback and Instructional Systems for Balance Rehabilitation Applications," Ph.D. Thesis. University of Michigan, 2012.

[7] J. W. Burke, M. D. J. McNeill, D. K. Charles, P. J. Morrow, J. H. Crosbie, and S. M. McDonough, "Optimising engagement for stroke rehabilitation using serious games," *The Visual Computer*, vol. 25, no. 12,

pp. 1085–1099, Aug. 2009, ISSN: 0178-2789. DOI: 10.1007/s00371-009-0387-4. [Online]. Available: http://www.springerlink.com/content/u623543wv9m0572l/.

[8] D. Fitzgerald, N. Trakarnratanakul, B. Smyth, and B. Caulfield, "Effects of a wobble board-based therapeutic exergaming system for balance training on dynamic postural stability and intrinsic motivation levels.," *The Journal of orthopaedic and sports physical therapy*, vol. 40, no. 1, pp. 11–9, Jan. 2010, ISSN: 0190-6011. DOI: 10.2519/jospt.2010.3121. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/20044704.

[9] M. Swan, "Sensor Mania! The Internet of Things, Wearable Computing, Objective Metrics, and the Quantified Self 2.0," *Journal of Sensor and Actuator Networks*, vol. 1, no. 3, pp. 217–253, 2012, ISSN: 2224-2708. DOI: 10.3390/jsan1030217. [Online]. Available: http://www.mdpi.com/2224-2708/1/3/217.

[10] P. B. Shull, W. Jirattigalachote, M. A. Hunt, M. R. Cutkosky, and S. L. Delp, "Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention," *Gait & Posture*, vol. 40, no. 1, pp. 11–19, 2014, ISSN: 0966-6362. DOI: http://dx.doi.org/10.1016/j.gaitpost.2014.03.189. [Online]. Available: http://www.sciencedirect.com/science/article/pii/S0966636214002872.

[11] H. Peoples, T. Satink, and E. Steultjens, "Stroke survivors' experiences of rehabilitation: a systematic review of qualitative studies.," *Scandinavian journal of occupational therapy*, vol. 18, no. 3, pp. 163–71, Sep. 2011, ISSN: 1651-2014. DOI: 10.3109/11038128.2010.509887. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/20701431.

[12] K. Motoi, S. Tanaka, Y. Kuwae, T. Yuji, Y. Higashi, T. Fujimoto, and K.-i. Yamakoshi, "Evaluation of a wearable sensor system monitoring posture changes and activities for use in rehabilitation," *Journal of Robotics and Mechatronics*, vol. 19, no. 6, p. 656, 2007.

[13] G. Chelius, C. Braillon, M. Pasquier, N. Horvais, R. P. Gibollet, B. Espiau, and C. Azevedo Coste, “A Wearable Sensor Network for Gait Analysis: A Six-Day Experiment of Running Through the Desert,” *IEEE/ASME Transactions on Mechatronics*, vol. 16, no. 5, pp. 878–883, Oct. 2011, ISSN: 1083-4435. DOI: 10.1109/TMECH.2011.2161324. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5975220.

[14] A. Karahanoglu and C. Erbug, “Perceived Qualities of Smart Wearables: Determinants of User Acceptance,” in *Proceedings of the 2011 Conference on Designing Pleasurable Products and Interfaces*, ser. DPPI ’11, New York, NY, USA: ACM, 2011, 26:1–26:8, ISBN: 978-1-4503-1280-6. DOI: 10.1145/2347504.2347533. [Online]. Available: http://doi.acm.org/10.1145/2347504.2347533.

[15] J. Rosenthal, N. Edwards, D. Villanueva, S. Krishna, T. McDaniel, and S. Panchanathan, “Design, Implementation, and Case Study of a Pragmatic Vibrotactile Belt,” *IEEE Transactions on Instrumentation and Measurement*, vol. 60, no. 1, pp. 114–125, Jan. 2011, ISSN: 0018-9456. DOI: 10.1109/TIM.2010.2065830. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5659690.

A Auditory Feedback for Slacklining

(a) Slackline (tubular, nylon, 1 inch width) (b) Slackline walking in Winter

Figure A.1: Slacklining

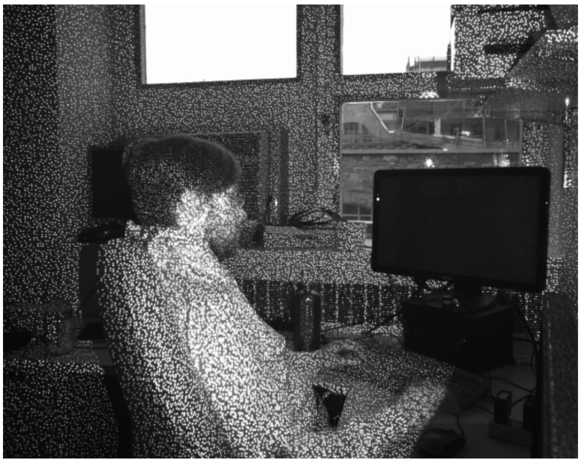
We implemented a proof-of-principle auditory balance feedback system for slackline learning, using a Microsoft Kinect to sense postural parameters. (ToDo: *At the time, it seemed like a good idea*)

Since motor learning and motor rehabilitation are based on similar neural processes, our initial goal is to first support healthy users in a balance learning task and then transfer the lessons learned to the development of a support system for physical and occupational therapy in the elderly and stroke survivor populations. We chose slacklining, the act of balancing and moving on a tensioned webbing, as application example for a motor learning task difficult for healthy subjects.

Slacklining, shown in Fig. A.1, has been investigated for balance training [1] and rehabilitation purposes, also in the older population [2], and it has been found to help build core strength [3] and ability to focus [4]. Others have employed Slacklining to investigate postural control [5], biofeedback loops [6], balance recovery mechanisms [7].

Our aim here is to provide augmented feedback to people beginning slackline training. When learning to slackline, users are commonly advised to keep their trunk as perpendicular to the ground as possible, and to balance by bending their knees. Here, we developed a system that provides feedback on trunk tilt and knee bend angle, and tested the data acquisition process in-situ at a slacklining event.

The Microsoft Kinect [8], then recently introduced, is a color-depth camera combination (RGB-D) that aims at capturing a console user's posture for game



(a)

(b)

Figure A.2: Image of the structured light pattern as seen by the monochrome intensity camera, and a reconstruction of the depth information visualized in voxel form through the PCL.

control. [1] Widely available and low in cost, and the related software frameworks promised to possibly provide sufficient in measurement quality for our application and thus be an alternative to classical motion capture systems [9]. It quickly became popular rehabiliation research, for example to create rehabiliation games [10], [11], to guide physiotherapeutical movements [12] and to predict fall risk [13].

The hardware comprises an RGB image sensors for color pictures and a monochrome image sensor, both with 1280×1024 pixel resolution. The latter sensor serves as a depth sensor, in conjunction with a infrared laser illuminator that projects a structured light pattern onto the scene, as shown in Figure A.2a. The Primesense PS1080 SoC in the Kinect computes a depth map from the displacement of this pattern and provides it as a video stream at 30 frames per second with a spatial x/y resolution of 3 mm and a depth resolution of 1 cm at 2 m from the sensor. Figure A.2b shows a 3D reconstruction of a scene.

The Kinect hardware supplies RGB-D data, that is a color intensity image and a depth map, software running on a host computer fits a skeleton onto this data and tracks it through subsequent frames. At the time of development, two proprietary

[1] Since the release of this original model at the time of the prototype development described here, Microsoft has introduced the Kinect 2 model that use Time-of-Flight depth sensing instead of structured light, improving accuracy, but reducing precision **Wasenmuller2017**.

frameworks (OpenNI/NITE by hardware manufacturer Primesense and Microsoft's eventually released Kinect for Windows SDK) existed for this purpose. We chose the OpenNI/NITE stack, as it was supported on Linux and others found better performance for whole-body measurements (stand-up-and-go test) than the Microsoft approach [14] that relies[2] on pre-trained sitting postures.

The feedback synthesis and a visualization of the skeleton, shown in Figure A.4, were implemented in *SuperCollider* [15]. We used Osceleton[3] to interact with the OpenNI API and send skeleton positions to SuperCollider. The entire application, including the skeletal tracking, worked with little noticeable latency at 30 fps in real-time on a first-gen Intel Core i7 processor.

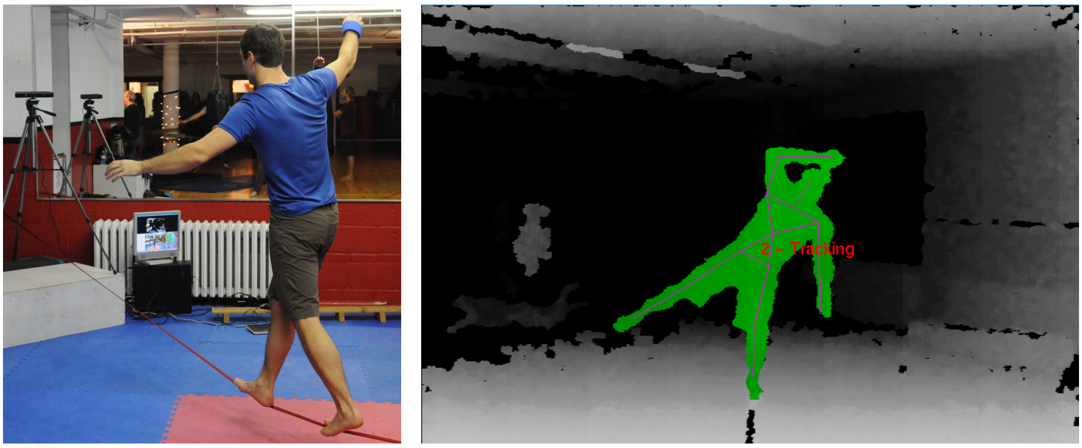
The baseline sound consists of two resonator banks, one high and one low in frequency, that are excited at a variable pulse rate. Changing the knee angle changes the pulse rate of the oscillators and also the resonance frequency of the banks, to become higher for increasingly bent knees, to finally approach a smoother sound. Tilting the upper body to the side changes the mixing ratio of the two resonator banks, and also the panning over the two stereo channels, to become a 'sharper' sound on the opposite side of the tilt direction. Correcting the posture thus has the effect of returning the sound to be balanced and smoother. A demonstration video can be found online.[4]

Commonly, the first objective for a beginner is to learn to stand stable on the line in all four different stances - left/right foot only, left/right foot in front, while keeping the upper body straight and perpendicular relative to the ground. As a first example we chose the single-legged stance, also favorable for the use of a single-camera optical tracking system since it involves almost no leg occlusions compared to the two-legged stance.

We chose sonification for our augmented feedback in a way that would make the user prefer to return towards ideal posture, providing knowledge of performance to the user on two key parameters: Knee bend angle (for the leg that is on the line, less is better) and deviation of the vertical axis of the upper body from the perpendicular (also less is better).

Example skeleton data of expert and intermediate slackliners was recorded during the gathering of the Canadian Slackline Association in January 2012 in Montreal.

[2] In the case of the first Kinect
[3] `https://github.com/Sensebloom/OSCeleton`
[4] `http://www.cim.mcgill.ca/~janlauff/slackson_demo_1.ogv`



(a) (b)

Figure A.3: Skeletal tracking of slackline trainees, setup and example skeleton overlaid on color coded depth image (darker = farther away).

Figure A.3a shows the setup used to record the data. The users where usually a bit further away from the camera to allow them to be captured in full.

The narrow distance range, in which an entire user body can be captured,[5] turned out to be a major drawback of the Kinect for this application. During the first data

[5]The depth map reconstruction works only in a certain distance range, as the patter becomes too bright and overexposes he camera for object that are too close, and becomes to weak for ones that are too far from the camera. Livingston et al. indicate an usable range for the Microsoft software implementation of 0.85-4 m, optimally 1.2-3.5 m in depth, with a field of view of 57 degrees horizontally × 43 degree vertically [16].

(a) (b) (c)

Figure A.4: Three one-legged slackline postures visualized in the SuperCollider frontend. The gauge in the upper left indicates the angle of the knee of the highlighted knee, the circle size of the joints the distance to the sensor.

collection, almost all feet were cut of in the recordings due of the high elasticity of the webbing used (15%) and the increased vertical space travelled by the participants. A major problem were self-occlusions, especially of the legs even with a favourable pose, and problems with ambient lighting such as in a room with daylight. Most of these problems seem to be also present with the Microsoft pose extraction, both algorithms use no per-user calibration and bone lengths vary from frame to frame, 2 cm for frontal and 5 cm for a 45 degree view of the Kinect w.r.t. the scene according to Kurillo et al. [17]. In our experience, bone lengths were even more unstable. Processing of Kinect data in real-time requires significant computing resources. The Kinect operates at 30 Hz, thus is fast enough for interactive, real-time use, but with at least 33 ms delay that may negatively affect closed-loop feedback systems for postural control.

Our goal here was to gain a qualitative understanding if the Kinect could be used for home balance training, for example in post-stroke recovery. We found that the tracking quality can be good, if and only if the conditions are favourable. Using more than one Kinect, e.g., with PCL and Skeltrak, could improve tracking quality. Evolving from early systems of marker-less tracking systems such as MeMoMan **Bandouch200**, modern deep-learning based approaches such as OpenPose **cao2017realtime**, **simon2017hand**, **wei2016cpm** make pose extraction from single intensity camera frames feasible and may be a suitable technology for replacing the Kinect in this system.

We presented an online, closed-loop augmented feedback slackline training system for the one-legged stance on a slackline. While the system was fully functional, technological challenges inherent to optical posture capture, such as dependence on lighting and computational complexity, severely limited its usability for a sport that is commonly performed outside. This exploration thus motivates the development of a body-worn instrumentation that is not dependent on external devices, and thus may be used in arbitrary situations.

B Vibrotactile Shoes (I 95%, S 80%)

(
words)

When we stand or walk, the feet are our contact point with the world. Much of our modern world is covered in hard, flat surfaces and the resulting sensory deprivation can lead to reduced balance ability. Plantar tactile perception, that is through the sole of the feet, appears to play an important role for this locomotion and postural coordination, e.g., by providing information about weight shift and environmental properties [1]. We may counteract this by providing strong plantar tactile stimuli [2], [3](jan: *spike sandal reference, maybe move to background section?*). Using vibrotactile haptics on the feet, we may uniquely relate to locomotion-specific events, for example by augmenting the ground impact during walking to provide information about floor texture [4].

Haptic communication through the feet may also offer advantages in applications for healthy users that are largely untapped in the state of the art. As a complementary modality to vision and hearing, haptics can be discreet and possibly ubiquitous. Among possible body locations for wearable haptic interfaces [5], [6], the feet stand out as fairly dexterous and exhibiting a higher resolution for haptic perception than most other body parts [7]. This makes them good candidates for interaction when the hands are occupied, as, for example, with the pedals of a car or foot switches in medical and industrial settings. This makes the sole of the foot a promising location both for placing rehabilitation wearables [8], [9] and navigation aids [6].

The bottom of the foot represents perhaps the most difficult and least studied place to render vibrotactile surfaces. Motion can mask haptic perception anywhere on the body, but the large forces involved in walking in combination with continuously changing coupling of the actuators throughout the gait cycle place special requirements on mechanical robustness and body-actuator coupling. Here, we investigate wearable haptic displays that incorporate multiple actuators and their intentful placement on anatomical features to facilitate discrimination by their users [10]. To enable interactive and rehabilitation applications, we also want to capture relevant metrics about the user's posture and gait. Our fully wearable approach can work without stationary equipment, embedded into regular footwear, in order to gain understanding of the specific challenges that come with a real-world system.

B.1 Background: Plantar Vibrotactile Systems

Foot-based wearables have long been explored, with most platforms designed for sensing-only functionality. Early examples include the pressure-sensing insole presented by Pedotti et al. [11] and the "expressive footwear" by Paradiso et al. that combined pressure, motion, and distance sensors with real-time data analysis. [12].

Systems with vibrotactile actuation emerged in more recent years, often for application in sensorimotor rehabilitation. For example, subthreshold noise rendered to the foot sole can improve balance control, as demonstrated by the insole of Priplata et al. [8] that used three vibrotactile actuators. Hijmans et al. [9] refined the design for real-world use and placed actuators at the four points that exhibit the highest pressure during gait and highest density of mechanoreceptors. We adopted these locations, for our system, as shown in Figure B.7.

Wantanabe et al. introduced a system for guiding users' walking pace, giving real-time vibrotactile feedback on steps sensed through pressure sensors [13], indicating that their system could be used for sports coaching and rehabilitation. Similar technology has been proposed to address gait freeze in patients with Parkinson's disease, for example by Winfree et al. [14], [15]. Papetti et al. introduced shoes with high bandwidth actuators able to synthesize the texture of virtual ground surfaces [16], and Zanotto et al. developed a fully wearable variant of the concept [17], again aimed at aiding Parkinson's disease patients.

An early study of foot-based tactons was presented by Velázquez et al. [18]. Comparing two custom insoles, one with 16 and one with four independent actuators, they found slightly better recognition rates for directional patterns presented on the latter. Subsequent experiments evaluated tactons for pattern recognition, emotion recognition and language learning. However, all experiments were conducted sitting, with the foot in the air and the insole strapped to the feet. Ménélas and Otis presented a serious game for learning six waveform-based tactons rendered on their custom shoe platform, in which the users walk through a maze on a smartphone [19].

Most recently, in a Wizard of Oz experiment, Meier et al. compared tactons for navigation cues on the different body locations proposed in the literature, namely foot (insole, shoe, sock), hip, and wrist [6]. For the shoe, they replicated the prototypes described by Velázquez et al. [18] and reported that compared to a visual-only condition, tacton delivery to the foot was successful in reducing stress and errors.

B.1.1 Tactile Perception on the Feet

Kennedy, P. M., & Inglis, J. T. (2002). Distribution and behaviour of glabrous cutaneous receptors in the human foot sole. The Journal of Physiology, (2002), 995–1002. https://doi.org/10.1013/jphysiol.2001.013087 [20]

Thompson, C., Bélanger, M., & Fung, J. (2011). Effects of plantar cutaneo-muscular and tendon vibration on posture and balance during quiet and perturbed stance. Human Movement Science, 30(2), 153–71. https://doi.org/10.1016/j.humov.2010.04.002 [21]

Gu, C., & Griffin, M. J. (2011). Vibrotactile thresholds at the sole of the foot: effect of vibration frequency and contact location. Somatosensory & Motor Research, 28(3–4), 86–93. https://doi.org/10.3109/08990220.2011.622493 [22]

Hennig, E. M., & Sterzing, T. (2009). Sensitivity mapping of the human foot: thresholds at 30 skin locations. Foot & Ankle International, 30(10), 986–91. https://doi.org/10.3113/FAI [23]

B.1.2 Single-Actuator / Stationary Systems

B.1.3 Multi-Actuator Shoes and Insoles

Grid [24]–[26] SoleSound [17], [27] Placement [9][28]

B.1.4 Spatiotemporal Patterns: Tactons

Brewster and Browns proposed “tactons”, tactile icons that are structured, abstract messages encoding information into frequency, amplitude and duration [29]. Tactons, defined by Brewster and Brown as “structured vibrotactile messages which can be used for non-visual information presentation” [29], have been studied on a variety of body parts and less so on the feet. With respect to tacton delivery using wearables with multiple haptic actuators, studies by Lee and Starner [30] and Brunet et al. [31] showed promising recognition rates with tactons at the wrist. These studies serve as inspiration for the foot-based tacton design outlined in this paper.

Aside from placement, the primary parameters of vibrotactile actuators that we can control are the amplitude (some previous systems used groups of actuators to compensate for the lack of independent control of amplitude and frequency of ERMs [32], [33]), and for certain actuators, spectral content (i.e., frequency and waveform) of the input signal. Spatiotemporal patterns on actuator arrays, optionally containing variations in these signals, have been explored by several researchers [30], [31], [34]. However, few have explored such interfaces on the feet, Magaa et al. [35] being one of the few exceptions.

At least in localization tasks, increasing the number of tactors of the display does not necessarily lead to superior perceptual performance. Rather, using anatomical points of reference when positioning the tactile display seems to be effective [36], [37]. There is little prior work on multi-tactor displays that are aligned with anatomical landmarks.

Gait Analysis

Howell, A. M., Kobayashi, T., Hayes, H. A., Foreman, K. B., & Bamberg, S. J. M. (2013). Kinetic gait analysis using a low-cost insole. IEEE Transactions on Biomedical Engineering, 60(12), 3284–3290. https://doi.org/10.1109/TBME.2013.2250972 [38]

Paradiso Shoe used for Gait parameters (Instrumented shoes with wireless capabilities demonstrated the feasibility of walking parameters computation such as heel-strike, toe-off, foot orientation and position) [39]

Walking Rehabilitation

Split-Belt Training [40] [41]

B.1.5 Gait Cueing / Parkinson's

Otis, M. J. D., Ayena, J. C., Tremblay, L. E., Fortin, P. E., & Mnlas, B. A. J. (2016). Use of an enactive insole for reducing the risk of falling on different types of soil using vibrotactile cueing for the elderly. PLoS ONE, 11(9), 1–26. http://doi.org/10.1371/journal.pone.0162107 [42]

[43]

B.1.6 Environmental Synthesis

Phantom Slippers [44]

B.2 Goals and Requirements

The literature review indicated promise for foot-based tactons, both while the user is stationary and during locomotion. Little preliminary work exists on self-contained, multi-actuator wearable devices. This motivated our anatomy-adapted and gait-adapting system design, build on a novel actuator integration robust enough to withstand the forces of gait. During the development of our prototype, the following challenges involved in a successful design became apparent.

B.2.1 Personas

We formulated set of personas encompassing several envisioned use cases. Personas are profiles for archetypal users, each profile representing a composite of a subpopulation of users [45].

(ToDo: *redo as table, pictures, credit, extend*)

Ferne

Age: 71, Occupation: Retired, Comfort with technology: 1/5, Active Life: 3/5 Ferne is a retired grandmother who stays active by going for a long walk every evening. A few years ago, however, Ferne fell in her kitchen and had to undergo hip replacement surgery on her left leg. After a long recovery period, she began to resume her daily walks around her neighbourhood. However, she recently began experiencing some discomfort in her good leg whenever she returns home from her walks. Ferne's doctor believes she has been placing too much pressure on her right leg, and has recommended that she routinely changes her cadence during her walks. Ferne wants to follow his advice, but forget to do so as soon as she slips into her habitual gait.

Paul

Age: 32, Occupation: Graduate Student, Comfort with technology: 3/5, Active Life: 1/5 Paul was involved in a biking accident last year, and sustained multiple fractures to his leg. His bones took several months to heal, after which his doctor referred him to a rehabilitation clinic. Paul's insurance as a graduate student will only cover one session per week at the rehab clinic. However, Paul feels that he is not progressing fast enough, and wishes he could do more to speed his recovery at home without running the risk of hurting himself any further.

Justine

Occupation: Graphic Designer, Age: 24, Comfort with technology: 5, Active Life: 2/5 Justine is a freelance graphic designer who loves video games. She enjoys her job, but the nature of her employment means she often works extensive hours. After a long day of work, Justine likes to relax by playing video games at home on one of her many consoles. However, Justine feels that between her job and favourite pass-time, she is not spending as much time as she should outdoors. Although she sometimes plays games on her smartphone in the park, she often feels the experience is neither as immersive nor engaging as when she uses her Kinect or Wii Fit at home.

Damon

Age: 39, Occupation: Chef, Comfort with technology: 3/5, Active Life: 4/5 Damon believes that seeking new culinary experiences is crucial to his success as chef. As a result, he tries to take a trip to a new part of the world every year. When traveling, Damon likes to immerse himself in a new city by walking around for hours on end with no destination in mind, trying out local foods as he goes along. Unsurprisingly, this has landed him in trouble a few times when he found himself completely lost in a city whose language he could not speak. Although Damon has a smartphone, he typically prefers not to use it on his journeys so as to not incur expensive fees. Damon wishes he could somehow maintain a general of direction while wandering in a new city, without having to constantly refer to maps, which he finds rather cumbersome.

According to our personas, we constructed a general-purpose prototype architecture that can function as a testbed for a number of applications, such as the following examples:

B.2.2 Intended Use Cases

B.2.3 Requirements

Integration and Body-Actuator Coupling

To support a range of typical human activity, our haptic mechanism must be functional within a regular shoe, so that we can conduct experiments in everyday scenarios, including sitting, standing and walking. This dictates constraints with regard to space, robustness, and energy consumption. As with all tactile stimulation, effective coupling of the actuators to the body is critical. The actuators have to be strong enough to be perceived, at least for any application that involves conscious interaction with the system. Their coupling to the user's body should either be reliable, or at least quantifiable, so that experimentation can reveal the effectiveness of stimulation under different loading conditions, as during gait.

There are numerous variables influencing the coupling of the actuator to the foot, and it proved to be beyond the scope of our initial prototype design to control for most of these. At a high level, the main challenge in our design was efficient delivery of the haptic signal to the intended region of stimulation on the foot, while isolating the perception of vibration from actuators positioned elsewhere. Additional variables include actuator loading, and thus, the physical connection between foot and insole, choice of materials, and material/impedance junctions. These variables must be factored into the design. Ultimately, what is most important is the user's perception of the haptic stimulus, especially during natural, everyday activities.

With our haptic shoe platform, we seek to support applications in neurorehabilitation and to enable hands-free and non-visual human-computer interaction. Balance and mobility problems, as well as gait asymmetries, are frequently presented in neurological disorders such as stroke, spinal cord injury and polyneuropathies. A system incorporating enhanced somatosensory inputs at the sole can be used for gait rehabilitation involving implicit learning and promoting more even weight bearing [46], or provide support for training and rehabilitation of the sensorimotor system [8]. Further applications include customized notification delivery, for example, to individual clinicians in a hospital environment.

(Add Citation: *[47]*)

B.2.4 General Design

B.3 Validation of Placement: Prototype 1, Tethered

B.3.1 Vibrotactile Actuator Choice

(ToDo: *reword!*) We started the prototype development by surveying the available actuators. To this end, we obtained samples of different ERMs and LRAs. Further, we acquired a Midé QP-20 piezoelectric actuator, driven by a Texas Instruments DRV8662 board, several first-generation Tactile Labs Haptuators and one mkII Haptuator.

The Haptuators offer impressive strength and frequency response. The second generation Haptuator is indeed compact and potentially suitable for shoe integration, but its prohibitively high price point precluded us from considering it for our initial prototype. It is conceivable that this kind of actuator will continue to shrink in size and appear on the mass-market at a later point.

The piezoelectric actuator is flat (0.8 mm in depth) and relatively inexpensive. However, it is stiff and occupies a large surface area (50.8 mm × 25.4 mm), which makes it problematic to integrate multiple such actuators in an in-shoe device. Although piezo actuators could conceivably be mounted inside in a shoe, this would not be straightforward, and the high voltage requirements poses a safety concern.

Of the remaining options, we chose LRAs, as these are an emerging technology well suited to our application requirements: small, energy-efficient, minimal lateral vibration, and reasonably priced. (ToDo: *fill in the gap: describe where to find details (Section* ??*), link to LRA description in technologies.., rationale for choice?*)

B.3.2 Actuator Placement

Positioning the actuators at anatomical landmarks, such as the heel, is likely to help users locate the stimulus reliably(ToDo: *cite landmark paper [10], [36]*). This in turn requires us to place the actuator(s) in a manner that is adapted to the user, at the very least, tailored to a subset of shoe sizes.

We opted to build a prototype in-shoe mechanism directly, based on the prior qualitative evaluation of the available actuators.

Hijmans et al. [9] refined the design for real-world use and placed actuators at the four points that exhibit the highest pressure during gait and highest density of mechanoreceptors. (ToDo: *duplicated sentence*) (Add Citation: *[28]*)

These locations contain most crucial mechanoreceptors [20], and also receive the most pressure during gait. With our rehabilitation applications in mind, these locations are key to successful locomotion as the healthy gait process develops over these four points. (ToDo: *citation needed*) Thus, they are also likely to offer the best coupling to the foot when it is loaded. We adopted these locations, for our system, as shown in Figure B.7.

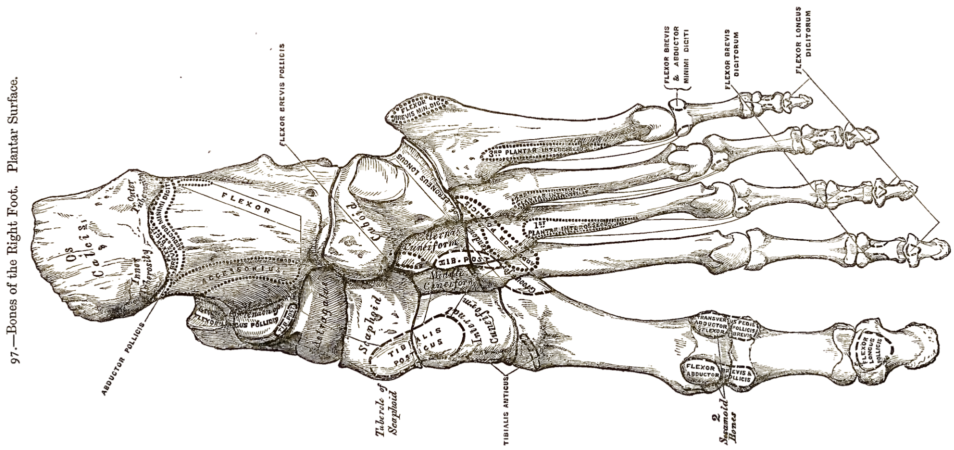
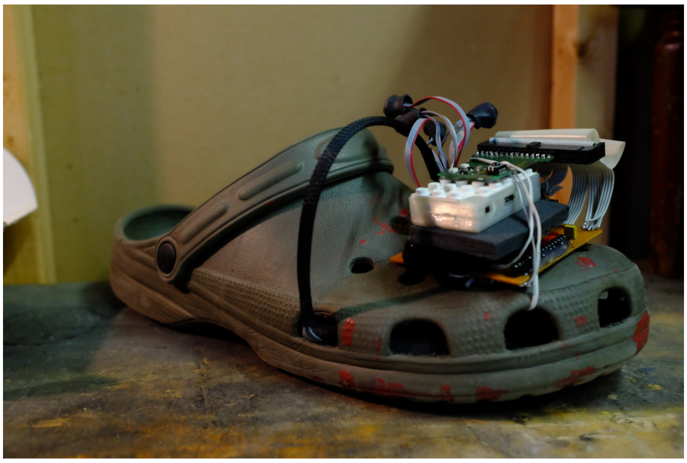


Figure B.1: Bones of the Foot, Plantar View (Figure adapted from [48])

B.3.3 Prototype Haptic Sandal

We use four pancake shaped linear resonant actuators (LRA), specifically, the Samsung DMJBRN0934AA (9 mm × 3.4 mm, $f_{res} \approx 205$ Hz, acceleration > 1G@100g). These are low-cost vibrotactile actuators that provide amplitude changes independent of frequency, and offer improved performance compared to the predominant eccentric rotating mass actuators. Their resonance frequency is tuned to be around 200 Hz, the region where the skin is most sensitive to vibration [49]. The LRAs are driven by one driver chip per actuator (Texas Instruments DRV2605L). For this stage of the project, we used effects from the library integrated in the DRV2605/DRV2605L chip.

The first prototype, shown in Figure B.2, was constructed as an insole for an inexpensive foam sandal, and based on four TI DRV2605L driver modules, an I2C



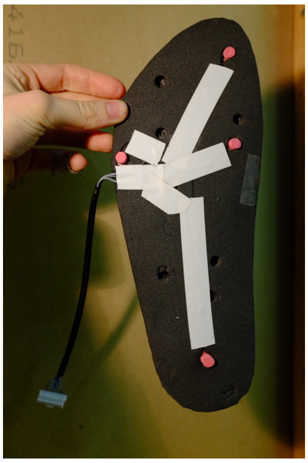


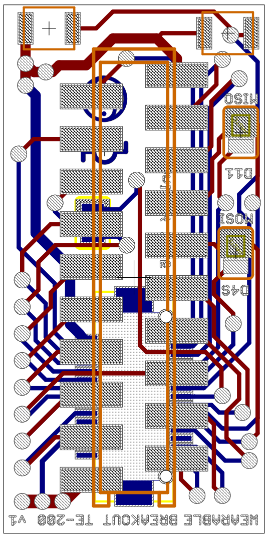
Figure B.2: Haptic sandal and insole, back side

bus switch and a $BRIX_2$ microcontroller. An untethered version of the haptic shoe will also require wireless communication for data logging, for which we hope to use the existing functionality of the $BRIX_2$ system.

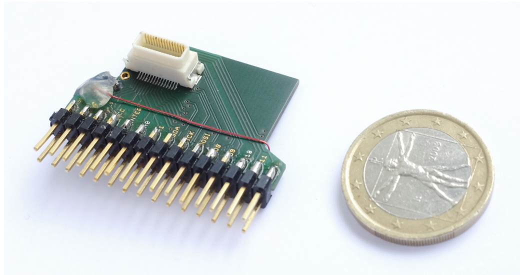
The insole consists of two sheets of foam, a thick sheet holding the actuators and spacers (pink), and a thin sheet as a covering. In conjunction with this insole, a size 13 shoe was found to provide a decent fit for users with shoe sizes 10-12 in our initial, static experiments. We placed four actuators at the first and fifth metatarsal head, the heel and the plantar side of the big toe [9], as described in the previous section. The placement of actuators within the insole was modified several times to align them well with the anatomical features of the first author.

$BRIX_2$ Wearable Breakout

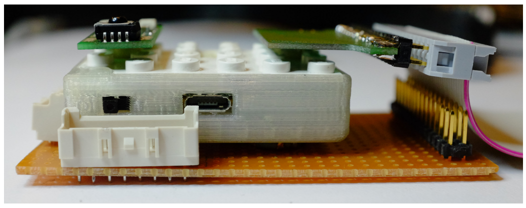
The $BRIX_2$ extension interconnect is based on a 30-pin Hirose DF17 connector, which provides almost many pins of the ATMega microcontroller to extensions. However, the connector's 0.5 mm pin pitch makes it mostly unsuitable for soldering by hand. Part of the official $BRIX_2$ extensions is a *breakout extension*, shown in Figure B.3 (b). It is meant to connect to a 30-pin IDC connector with a ribbon cable, which is quite large and means the extension protrudes significantly extends over the side of the side of the base module. While integrating a $BRIX_2$ into the initial shoe prototype (discussed in Section B.3) using the $BRIX_2$ breakout extension, we found that this setup did not result in a mechanically robust setup, as can be seen in Fig. B.3 B.3c. The cable and connector can easily lever the extension of the base controller.



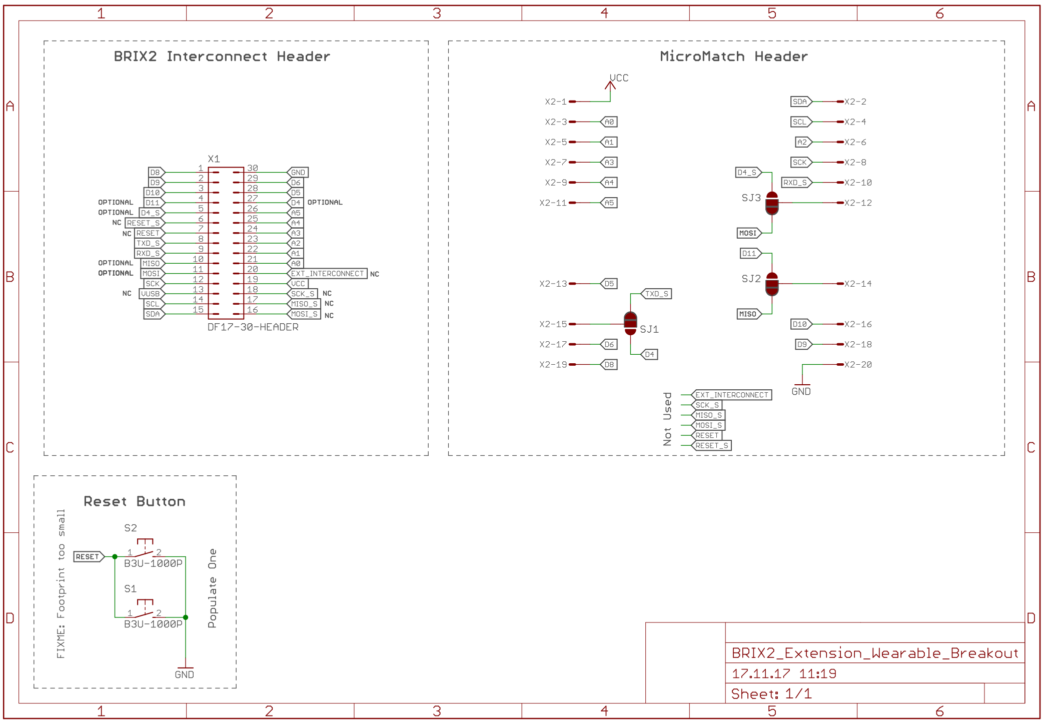
(a) Wearable Breakout Extension, PCB layout



(b) Original breakout extension



(c) Lever forces on right extension can not be held by DF-17 connector



(d) Wearable Breakout Extension: Schematic

Figure B.3: BRIX$_2$ Breakout Extensions

To address the need for custom connections to the $BRIX_2$ system We designed a "wearable extension" that maps a sub-set of most important extension pins to a 20-pin TE MicroMatch connector (MM-20).[1] We included a button to reset the microcontroller, which helps in situations where the automatic reset (through the DTR lines of the USB-Serial interface) of the Arduino IDE does not work. Table B.1 shows the pin mapping, Fig. B.3 B.3a the PCB design (jan: *no actual picture exists, build up module stayed in Montreal?*). (ToDo: *Application*)

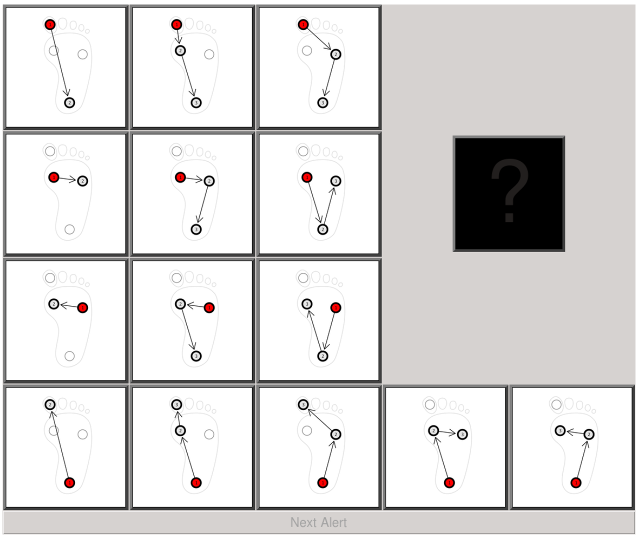
Table B.1: Pin map of $BRIX_2$ extension signals on wearable breakout extension header

Pin	**Signal**	**Pin**	**Signal**
1	`VCC`	11	`A5`
2	`SDA`	12	`MOSI` or `D4_S`
3	`A0`	13	`D5`
4	`SCL`	14	`D11` or `MISO`
5	`A1`	15	`TXD_S` or `D4`
6	`A2`	16	`D10`
7	`A3`	17	`D6`
8	`SCK`	18	`D9`
9	`A4`	19	`D8`
10	`RXD_S`	20	`GND`

This connector can be sourced as surface mount and through-hole version and is easily soldered by hand. It is an 1.27 mm pitch insulation displacement connector (IDC) and the connectors are squeezed onto the cable, so cables can be made easily by hand to fit using standard ribbon cable. MicroMatch connectors are more compact than normal 1.27 mm pitch IDC connectors, especially in component height. The extension has the regular footprint, which allows it to be placed on any of the three connectors, along with two other extensions to be placed on the base controller. Figure B.3 (a) shows the board layout. In the schematic, Figure B.3 (d), `NC` indicates that the pin is not broken out from the DF-14 to the MM-20 connector, and "optional" indicates that it shares its MM-20 pin with another DF-14 pin; a solder jumper can be set to choose.

[1] `http://www.te.com/usa-en/products/connectors/pcb-connectors/wire-to-board-connectors/ffc-fpc-ribbon-connectors/intersection/micro-match.html`

B.3.4 Tacton Discrimination: Sitting vs. Standing



(ToDo: *make this paper style*) We first conducted a small pilot study (n = 3), in sitting and standing condition. After a short training period, participants were asked to distinguish between 14 tactons, each consisting of a sequence of activations at two or three unique actuator locations. The pilot presented the activations using two different stimuli: a continuous buzz or three buzzes of the same total duration. Encouragingly, we found no difference for the two conditions or stimuli types, with pilot participants demonstrating near-perfect recognition rates. Anecdotally however, the actuator locations were hard to discern when the foot was not loaded. We developed the actuator insert described in Section B.3.3 to address this cross-coupling problem in the main experiment.

The goal of our initial experiment was to confirm the ability of users to learn and discern 14 different haptic patterns, over a set of four actuators in the in-sole. The experimental design was modeled on a similar study on multi-actuator haptic icons on the wrist by Lee and Starner [30].

To support the experiment, we developed firmware that can render arbitrary sequences of patterns, and an interface that presents training, test and trial sequences to the participants and records their answers. The related ethics approval, experiment instructions and interface are provided in Appendix B.

The spatiotemporal patterns used in the experiment consisted of sequences involving either two or three actuators, leaving out the top triangle patterns, which we found to be hard to distinguish in pilot experiments. Each pattern was either composed out of two or three single, longer pulses or three shorter ones of approximately the same length (from the DRV2605's library). The options we considered are illustrated in Figure **??**.

The participant was first exposed to the patterns in a pre-set order, then given a training phase wherein the graphical representation of each haptic pattern is revealed. This was followed by the main experiment, conducted with the participant both standing and sitting, during which each pattern was presented ten times in total. The participant must select the corresponding pattern without any feedback as to whether the response is correct. Finally, a shorter variant of the main experiment was conducted in which each pattern was presented only a single time.

The experimental results confirmed that 14 patterns can be learned and distinguished by all participants, both sitting and standing. Figure B.4 shows the reaction times for each participant: the reaction times stayed fairly consistent throughout the experiment per participant, but vary significantly between them. Recognition rates appear to be nearly perfect for all participants, and are thus not included in the plot.

B.3.5 Discussion

The initial research activities carried out to explore the feasibility of rendering discriminable haptic stimuli to the feet proved successful. Our efforts yielded several crucial insights, and suggested several promising directions for the continuation of this research.

From the qualitative and quantitative studies performed with stationary actuator arrangements, we identified several variables that may prove important to control in order to ensure reliable signal delivery. Most obviously, these include actuator coupling and its potential variability during the gait cycle.

(ToDo: *lead over*)

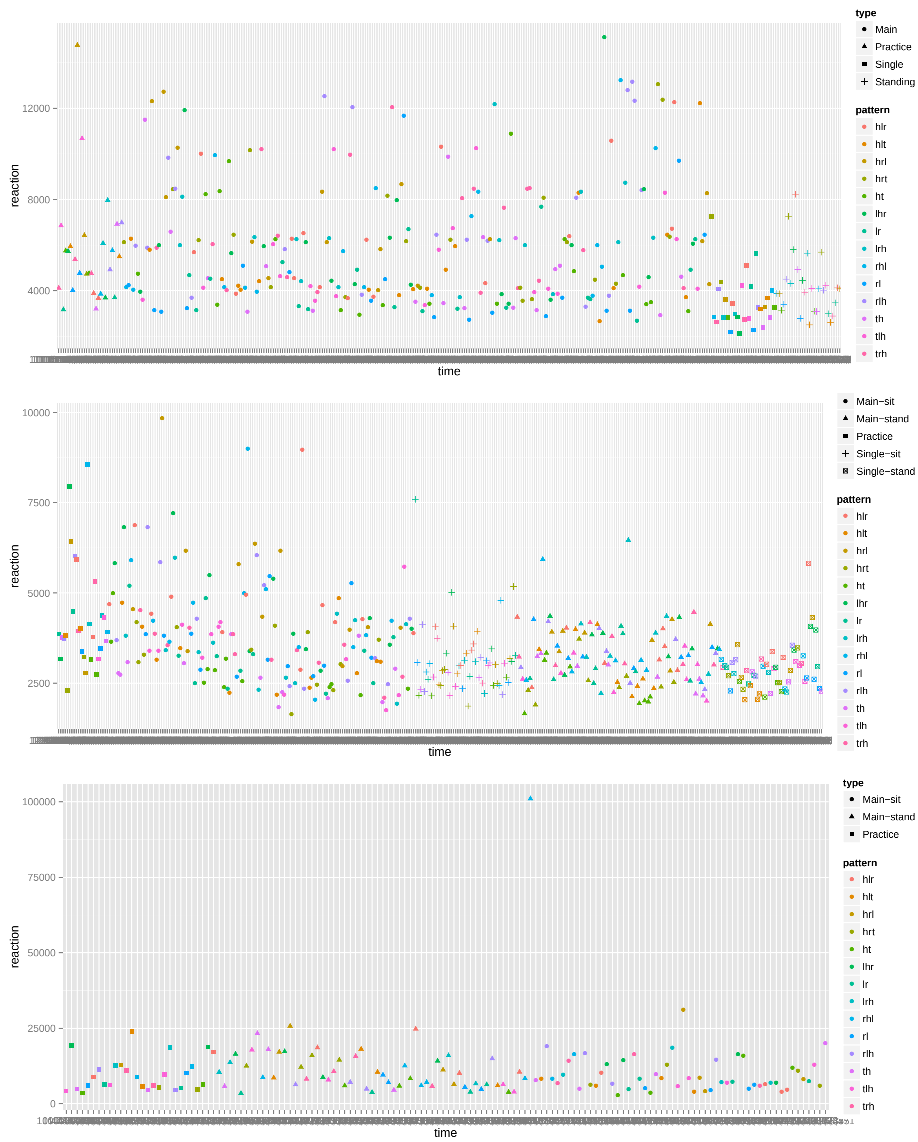


Figure B.4: Experiment results per participant: Reaction time for each pattern over the course of the entire experiment. The pattern are identified by their locations in sequence; “THLR” signifies **T**oe-**H**eel-**L**eft-**R**ight.

B.4 Improved Coupling and Integration for Locomotion: Prototype 2, Untethered

B.4.1 Goals and Requirements

B.4.2 Improved Actuator Integration

We conducted a number of experiments to establish how well different arrangements of foam and support materials, some shown in Figure B.6, work. Different foam, rubber, and silicone samples were tested. An analog ADXL320 accelerometer was held with fixed pressure to a stack of materials bonded to a constantly vibrating actuator. From the oscilloscope measurements, we observed certain variations of the actuator-generated vibration waveforms. The waveform is significantly modified by different materials, both in amplitude and shape, and also with the amount of loading. Figure B.5 shows a group of support material samples and one arrangement of the testing apparatus. However, it proved difficult to recreate the loading conditions of an actual foot, and impractical with this setup to measure the perceived strength of vibration. This suggested the importance of data collection under more realistic conditions in order to inform the design of an appropriate assembly of materials housing the actuators.

Figure B.5: One group of material samples, and an example accelerometer arrangement.

To explore the effects of different foams and actuator couplings in an insole-like substrate and with our feet, we developed a prototyping platform with exchangeable actuator modules, shown in Figure B.6.

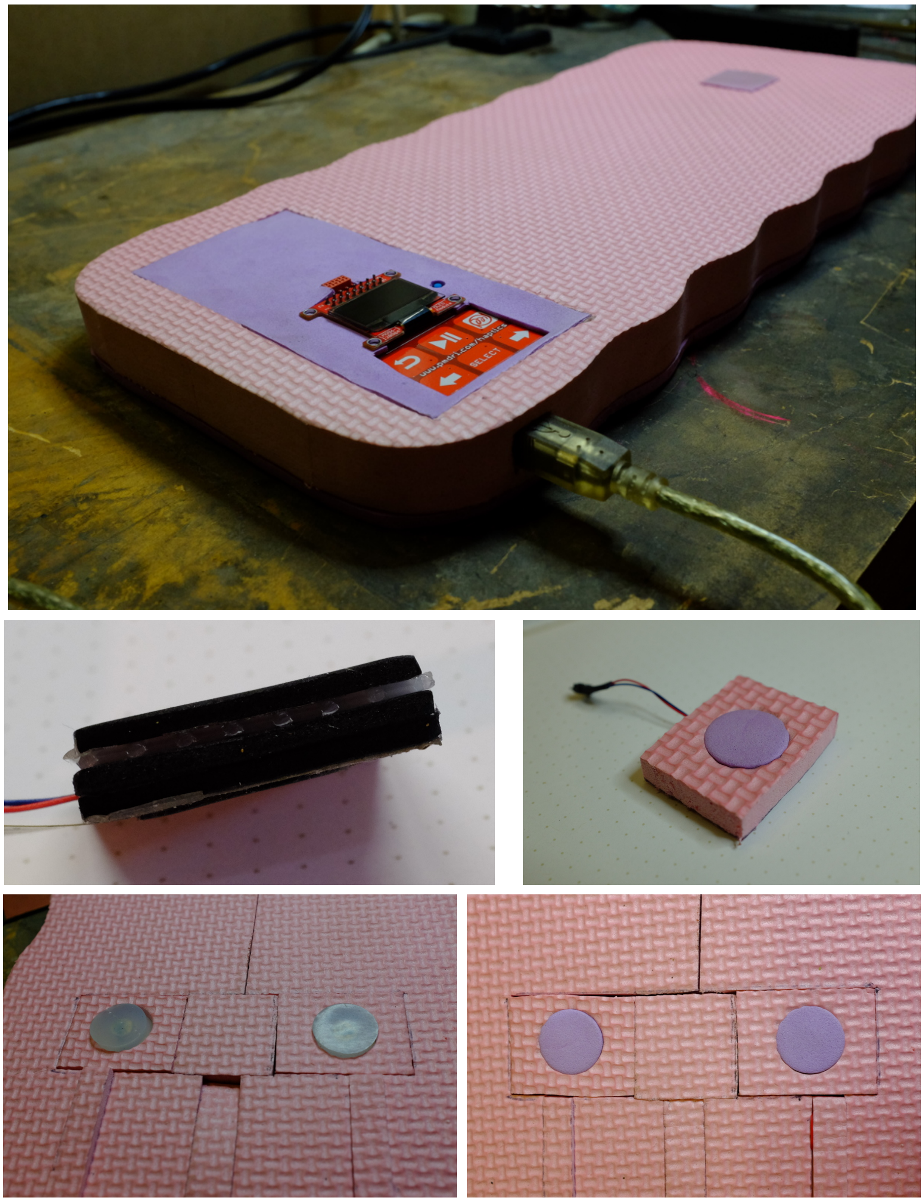


Figure B.6: Improve integration & actuator coupling, robustness: coupling testbed, inserts

In principle, the challenge is an impedance matching problem between the actuator and the plantar foot surface. The sole of the shoe functions similar to a baffle in a loudspeaker.

Dished aluminum inserts filled with silicone provide mechanical protection and work as a baffle for the vibration energy, guiding it to the user rather than into

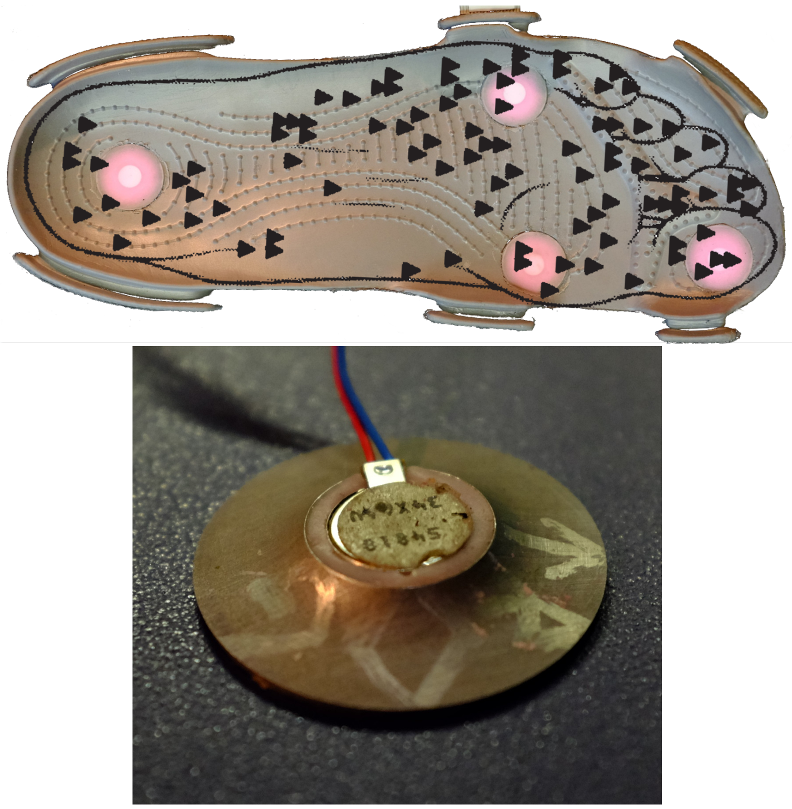


Figure B.7: Insole with actuator inserts and mechanoreceptor afferent units (adapted from Kennedy et al., 2002)

the sole and effectively decoupling the actuators from each other. The inserts are embedded in the shaped insole of a foam sandal (size US 10.5), shown in Figure B.8, which provides guidance to center the foot in the shoe. Figure B.7 shows the insole with the actuator inserts and the afferent units of the different mechanoreceptor types overlaid.

B.4.3 Untethered System Architecture

Sensors, Connectivity and Control

Underneath each of the four actuators are force-sensing resistors (Interlink FSR 402) to measure the loading at these key points. This enables us to measure weight shift over the user's base of support. The signals are conditioned for linearity with an op-amp circuit. (ToDo: *describe opamp conditioning somewhere*) On the side of the

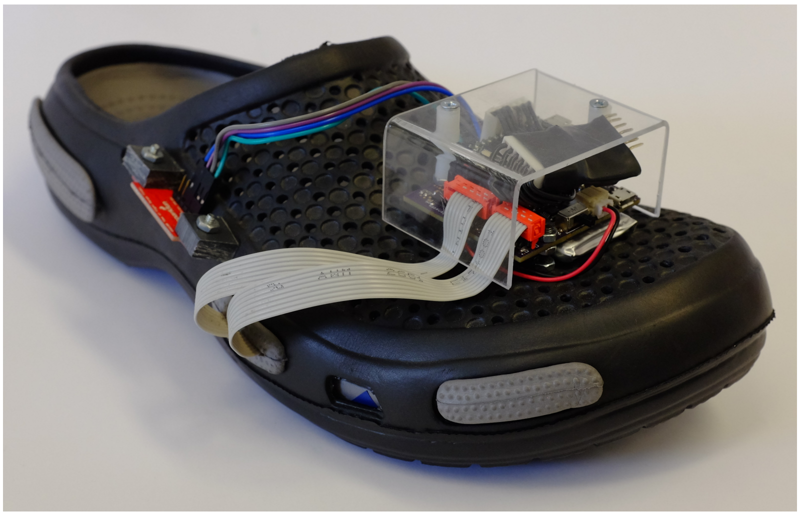


Figure B.8: Haptic Shoe Platform

shoe, we installed a Time-of-Flight based optical proximity sensor (ST VL6180X) providing absolute, millimeter-precision distance readings. This is used to measure step height and subsequently trigger playback of the tacton sequence when the shoe is very close to the ground, in anticipation of footfall, when the actuators are best coupled with the foot. Figure C.3 shows an excerpt of the proximity reading during walking and the threshold line.

(ToDo: *bluetooth latency depends on client, >= 7.5 ms* `https://drumpants.uservoice.com/knowledgebase/articles/483196-latency-and-lag`) A custom PCB combines the analog circuitry with a 32-bit microcontroller with 13-bit analog-digital converters (Teensy 3.2), an ultra-low power Bluetooth module for wireless communication (RedBear Labs BLE Nano), and charging/power management circuitry for lithium-polymer batteries that provide multiple hours of battery life. Finally, the board includes a digital readout 9-DoF motion sensor with integrated sensor fusion (Bosch BNO-055) for future applications. The individual components are described in detail in the modular toolkit chapter (ToDo: *detailed reference*). The electronics and battery are mounted on the front part of the shoe, as seen in Figure B.8.

An iOS or Android smartphone running an Evothings/Cordova application serves as the experiment controller, interface to the user, and stores both experiment data and received sensor readings into a remote database system over the wireless network

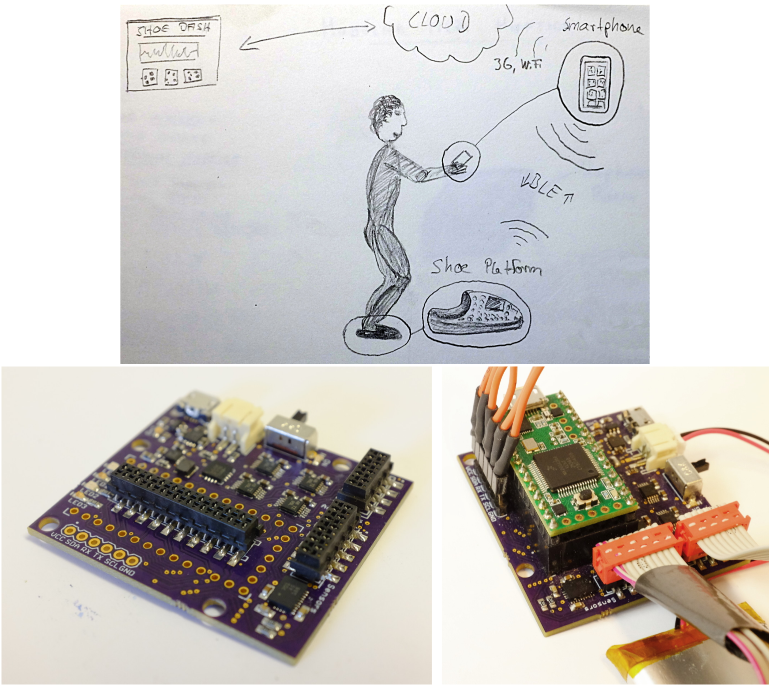


Figure B.9: Shoe Electronics

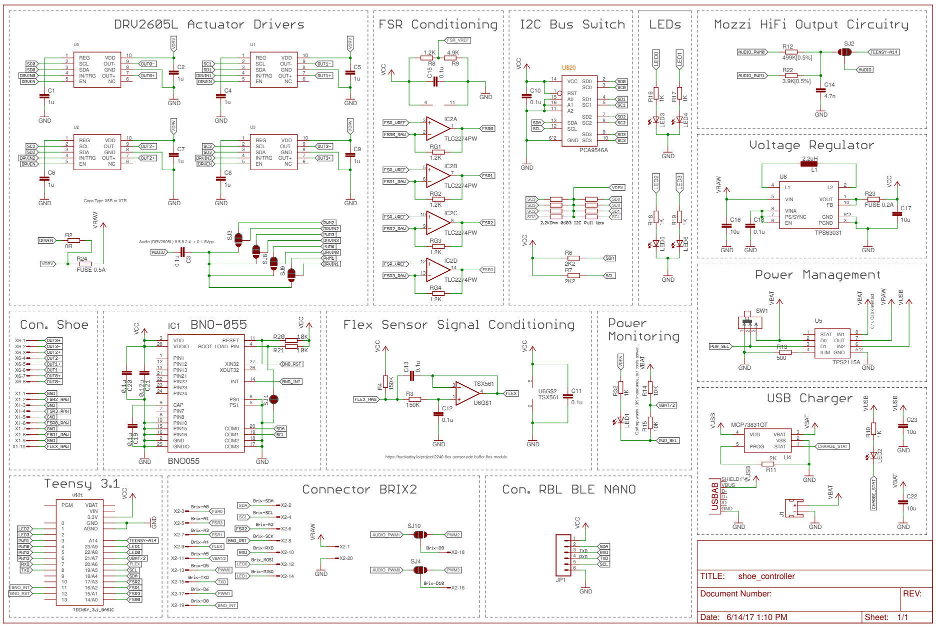


Figure B.10: Shoe Controller mk2 Schematics

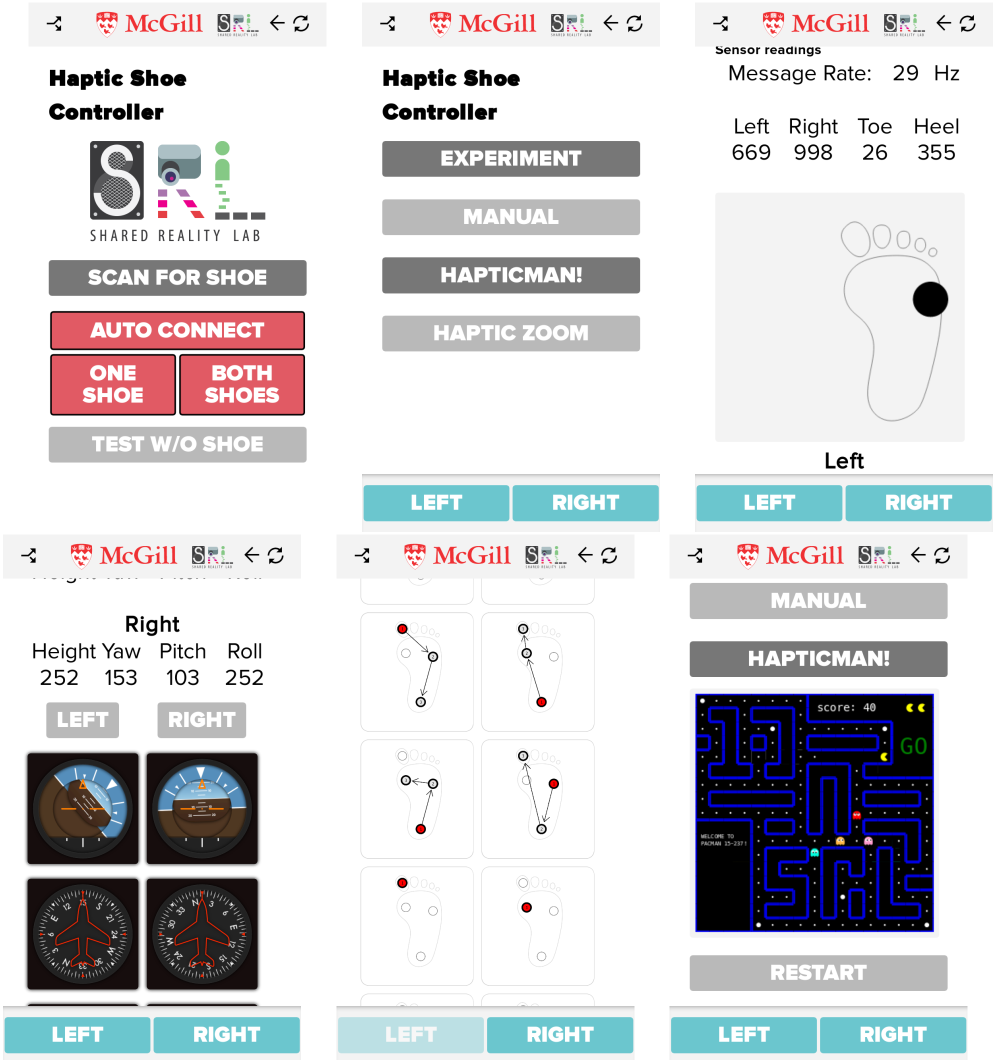


Figure B.11: Smart Shoe App (Cross-Platform)

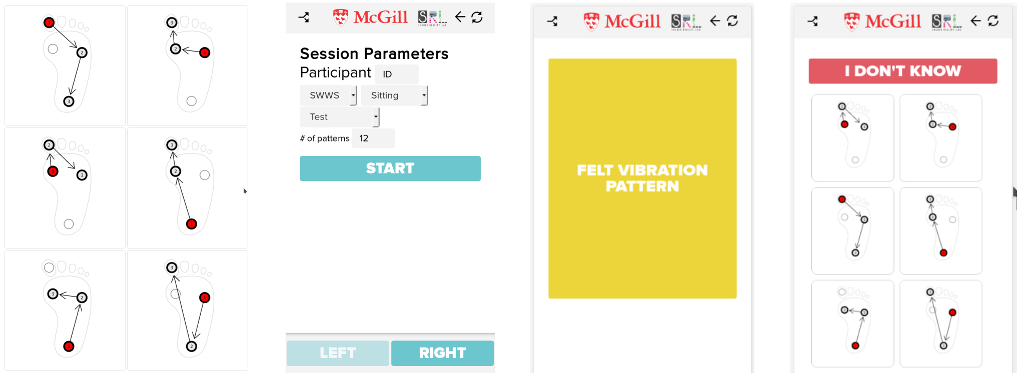


Figure B.12: Shoe App Views for the Tacton Study andMain experiment tactons, red circle indicates starting actuator location. Top left to bottom right: TRH, RLT, LTR, HLT, HRL, RHT

connection. This enables remote inspection of the data through a dashboard interface while the experiment is running and provides on-device buffering and offline support to ensure smooth operation even if the wireless data connection of the smartphone is interrupted. All time-critical operations, such as sensor readout and tacton playback, run locally on the shoe's microcontroller.

B.5 Discussion

Here, we present a custom-built, instrumented haptic insole that enables us to render tactile stimuli during gait and gain insight into relevant gait parameters, such as step height and weight distribution over the foot. Using a proximity sensor to detect footfall, the insole provides consistent triggering at a defined point in the gait cycle when the actuators should be best coupled to the sole of the foot.

C Vibrotactile Shoe Evaluation

The focus of this paper is to describe our system architecture and provide an initial evaluation of the system's performance as an information display. To this end, we measured the recognition rates six tactons, rendered to the foot sole, both during quiet stance and walking. The experiment also served to aid our understanding of which design aspects of the tactons are preferable for the anatomically and perceptually beneficial actuator placement locations we chose.

C.1 Tacton Discrimination: Standing vs. Walking

C.1.1 Introduction

C.1.2 Tactons

Lit Review

The purpose of the experiment, based on Lee and Starner's work [1], was to study tacton recognition rates with a prototype insole.

For the main experiment, we wished to compare recognition rates during quiet standing and during walking. Based on the promising results of the pilot, we hypothesized that participants would reach at least 90% accuracy during standing. We expected pattern distinction to be much harder during walking, due to the haptic artifacts generated from gait and varying contact between actuators and foot sole.

Tacton Design

As actuator location alone is a powerful tacton design dimension [2], we chose to keep frequency, wave shape and actuation length constant. The tactons we investigate here are sequences of buzzes of three different actuators in sequence, 250 ms in duration each. One entire tacton roughly fits into the time the foot is in contact with the ground, as described later (Section B.4.3 and Figure C.3). We subsequently denote tactons by the sequence of initials of actuator location, e.g., HLT would be Heel-Left-Toe. An example pattern is illustrated in Figure C.1.

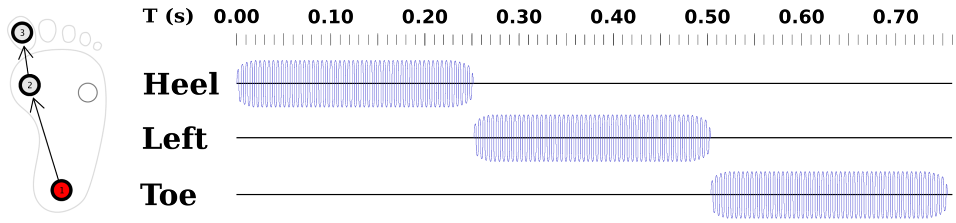
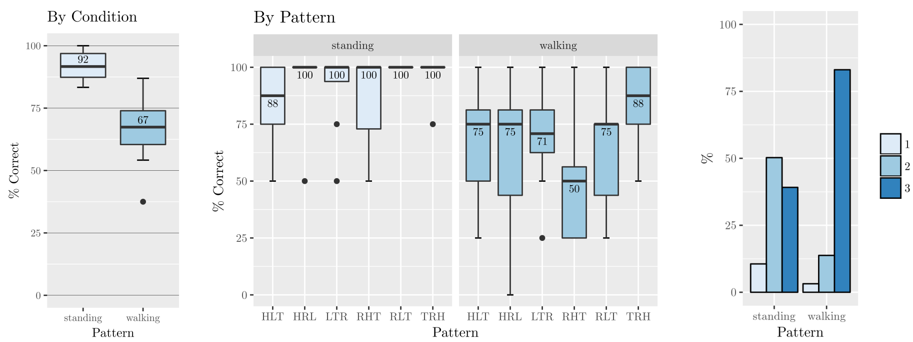


Figure C.1: Pattern HLT: *H*eel-*L*eft-*T*oe (HLT)

We chose six representative tactons, shown in Figure B.12, each composed of three consecutive vibrotactile stimuli. These included both patterns that have a traversal across the foot (TRH, HLT), as well as patterns that operate within a triangle (RLT, LTR, HRL, RHT). These patterns were selected in the hopes that they would offer insights with regard to rendering haptic illusions in the future, as all patterns have disparities in terms of direction (straight line or triangular) and area of effect (whole foot, upper half, or bottom half).

Eight male subjects (ages 19-27 years, mean 25.2 years) with self-reported shoe sizes between 10 and 11 participated in this experiment. All but one participant reported being right foot dominant. Participants were asked to wear the haptic shoe system on their right foot and hold a smartphone in their hand. While sitting, standing or walking, they were presented with one of six tactons and asked to select the tacton they perceived as soon as they were confident about identifying it correctly. Selection of the recognized tacton was performed by tapping on the associated pictogram, displayed on the smartphone screen, similar to Figure B.12 plus a “Don’t know” option. There was a random wait between 3 and 15 seconds between presentation of the tactons. The patterns were repeated up to three times within each trial. Total experiment duration was approximately 45 minutes.



(a) Recognition Rates (b) Recognition Rates by Pattern (c) Repetitions Needed

Figure C.2: Tukey boxplots for recognition rates and repetitions needed.

C.1.3 Study

Before the experiment, a manual test was performed to verify that the participants could feel all actuators and that the shoe was a proper fit. The experiment consisted of three sections: Introduction (seated, one set), training (standing, one set), and main (standing and walking, two sets each). A minimum three minute break was enforced in between each set to prevent sensory adaptation. During the experiment, participants wore headphones with pink noise adjusted to a volume level that prevented them from hearing the stimuli.

In the introduction, the system was explained to the participants, they gave informed consent and put on thin socks. Then, the tactons were presented indivdiually in combination with their respective pictograms. In training and main sections, participants were presented with a randomly chosen tacton, and asked to select the recognized tacton on the smartphone. For training, participants were given feedback to confirm a correct or inaccurate pattern selection.

For the main section, we divided the participants equally into two groups, either following a Walk-Stand-Stand-Walk (WSSW) or a Stand-Walk-Walk-Stand (SWWS) combination. This was done to balance out potential learning effects. In the standing condition, participants were asked to stand straight and distribute their weight evenly across both feet. In the walking condition, participants were instructed to walk around a circular hallway at a normal pace, in a counterclockwise direction. The haptic shoe was set to hold off playback of the tactons until the measured step height fell below a threshold of 60 mm, 10 mm above the baseline during standing (chosen

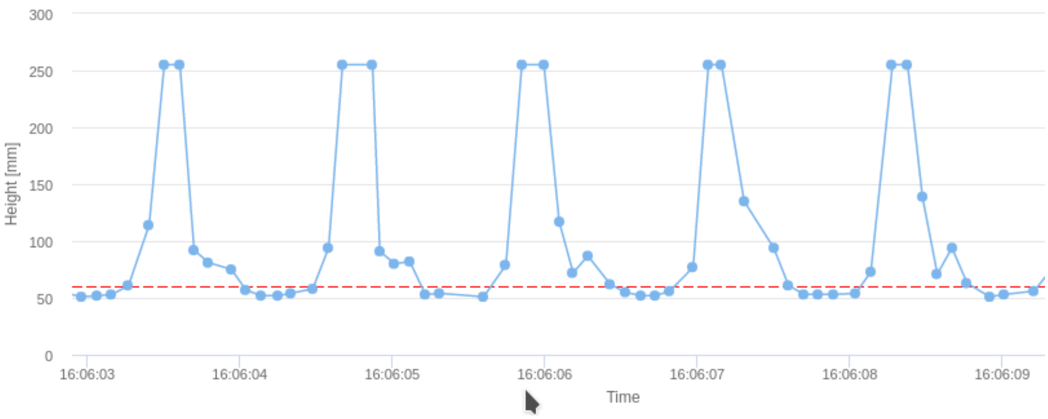


Figure C.3: Example step height plot, during walking, red line indicates 60 mm threshold for pattern delivery. The low temporal resolution was subsequently fixed; as the trigger point was computed locally on the shoe actuation timing was not affected by this.

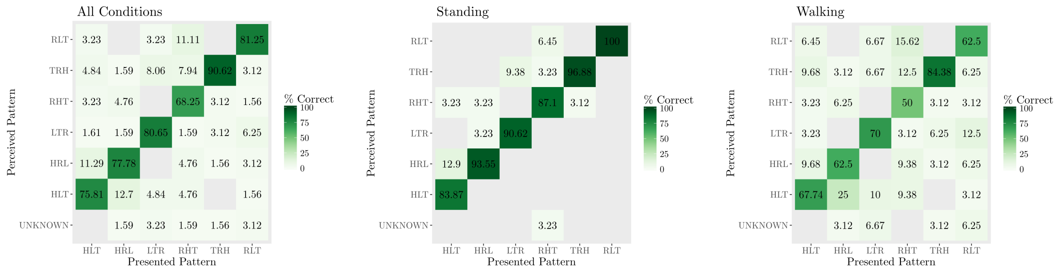
empirically, also see Figure C.3). This was done to ensure full contact with the ground and thus hopefully optimal contact between all actuators and the sole of the foot.

All participant answers, sensor readings and reaction time were recorded. After the experiment, participants completed a questionnaire to gather additional information, such as preferred patterns and whether they found the tactons easy to perceive.

C.1.4 Results

Tukey boxplots for the recognition rates can be seen in Figures C.2a-b; Figure C.5 shows all data points.

Even with little training, recognition rates during standing are good (mean = 92.12%, SD = 6.53). As expected, recognition rates were significantly lower during walking, but still above chance (mean = 66.26%, SD = 15.78, confirmed with paired t-test: $p < 0.001$). Only four presentations were missed in total, and seven presentations were answered with “Dont’ know”. Even though more tacton repetitions were used in this condition (see Figure C.2c), participants most often waited for all three presentations before entering their decision. Most participants reported the system to be comfortable, but all reported the walking condition to be much harder than the others.



(a) Standing and walking combined (b) Standing (c) Walking

Figure C.4: Confusion Matrices

Anectodally, most participants reported their preferred patterns to be those that included the extremes of the foot, and that spanned the whole length of the foot (TRH, HLT). This is in agreement with the better recognition rates seen in the confusion matrices of Figure C.4. As hypothesized, the results confirm that users can generally distinguish between the six patterns during standing and walking, and can do so somewhat during walking.

C.1.5 Discussion

C.1.6 Conclusion and Outlook

We presented a user study evaluating the perception of six tactons during standing and walking using a custom, gait-aware haptic shoe. Future work should incorporate the data of the other sensors to allow a better analysis of recognition failures; i.e., whether these are caused by insufficient contact between foot and actuator. A larger, more diverse group of participants should be tested to derive more generalizable results and gain more understanding of why certain tactons are recognized better than others. This will require the construction of multiple sizes of shoes ideally built around stronger haptic actuators to improve the signal-to-noise ratio during walking.

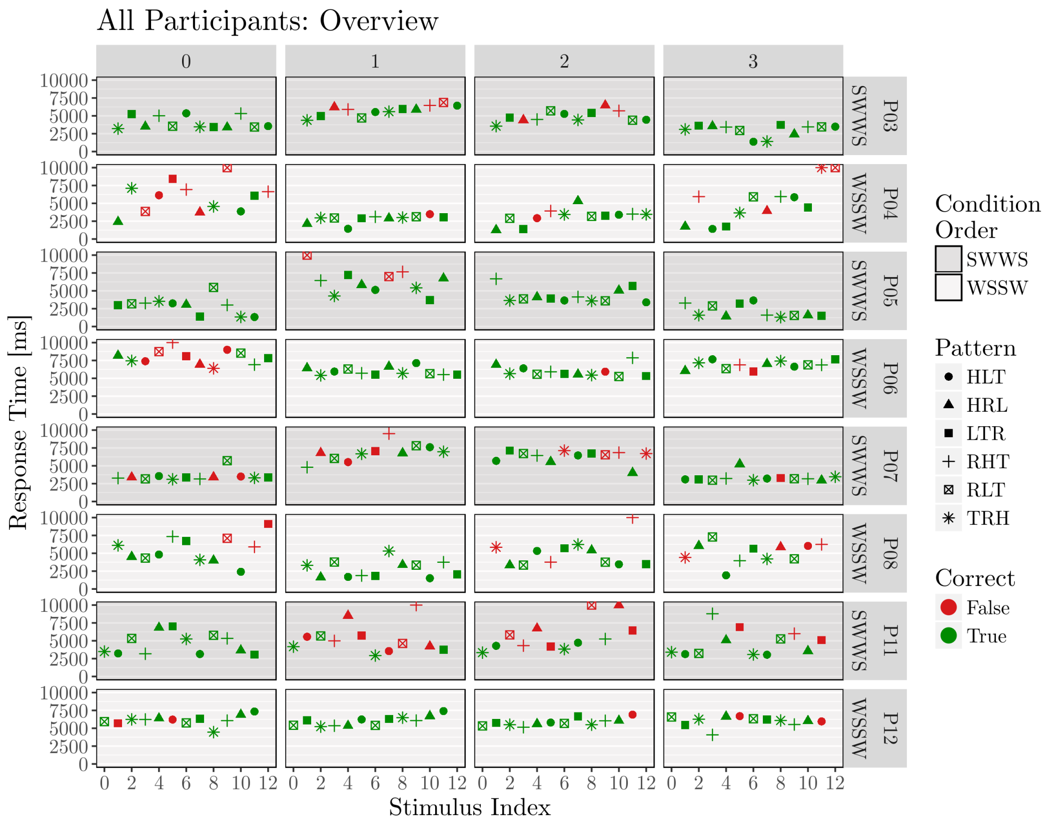


Figure C.5: All presented tactons and answers. Three response time outliers were clipped at 15 seconds.

C.2 Effects of different feedback modalities on radial menu selection tasks

C.2.1 Introduction

Finally, we believe that foot-based haptic interfaces have potential for creating bi-directional hands-free interfaces.

Foot-based interfaces are commonly used in environments where hands and eyes are occupied. They excel at simple control task and assisting the hands [3], such as in control of automobiles, aircraft, or tool speed. As computing increasingly becomes mobile, we are interested in addressing the question of whether interfaces based on commodity sensor technology, integrated into footwear, can be controlled effectively without visual feedback?

Depending on the environment, relying on the auditory channel may not be feasible due to noise or existing signals, such as in medical or industrial settings. Thus, our interest here is to investigate if vibrotactile feedback can provide comparable performance to auditory feedback. Similarly space and user movement may be heavily constrained, as in surgical theatres.

Here, we investigate the selection of items from a radial menu by foot rotation, seeking to determine whether localized haptic feedback serves as a viable alternative to both visual and auditory feedback, based on user performance. Fig. C.6 illustrates our 1-D radial menu selection task. To make menu selections, participants relied on natural proprioceptive feedback, and possibly, visual tracking of their own foot position. In certain conditions, they were also provided with augmented auditory or vibrotactile feedback. (jan: *edited previous paragraph again for figure reference, please proofread.*) (jan: *actually, re-reading the paper: they did not propose this, so I think it's stronger to leave out. we're citing the paper below. their paper unfortunately does not report any actual numbers, just statistical differences.*) Here, we compare a simple click sound to two vibrotactile stimuli, the latter rendered on a multi-actuator insole to investigate whether localized feedback can provide benefits at higher menu item counts.

C.2.2 Related Work

Research on foot-based wearables ranges from early instrumented footwear [4], [5], to more recent exploration of haptic actuation, rendering immersive virtual environments [6], [7], supporting rehabilitation [8] and navigation [9], [10]. Foot haptic feedback can also be used to improve gesture interfaces for people with severe motor dysfunction [11]. Velloso et al. provide an extensive literature review of diverse foot interfaces and compared different foot interactions in 1D and 2D based on Fitts' law tasks [3]. Scott at al. presented a system to classify foot-gestures, such as ankle rotation, from a mobile phone in the pocket [12].

Hatscher et al. identified foot rotation and relative as opposed to absolute movement be a suitable gesture for controlling 1-D parameters [13]. Grane et al. evaluated the advantage of haptic feedback in rotary menu selection tasks by adding different haptic texture effects. They found that a combination of visual and haptic feedback

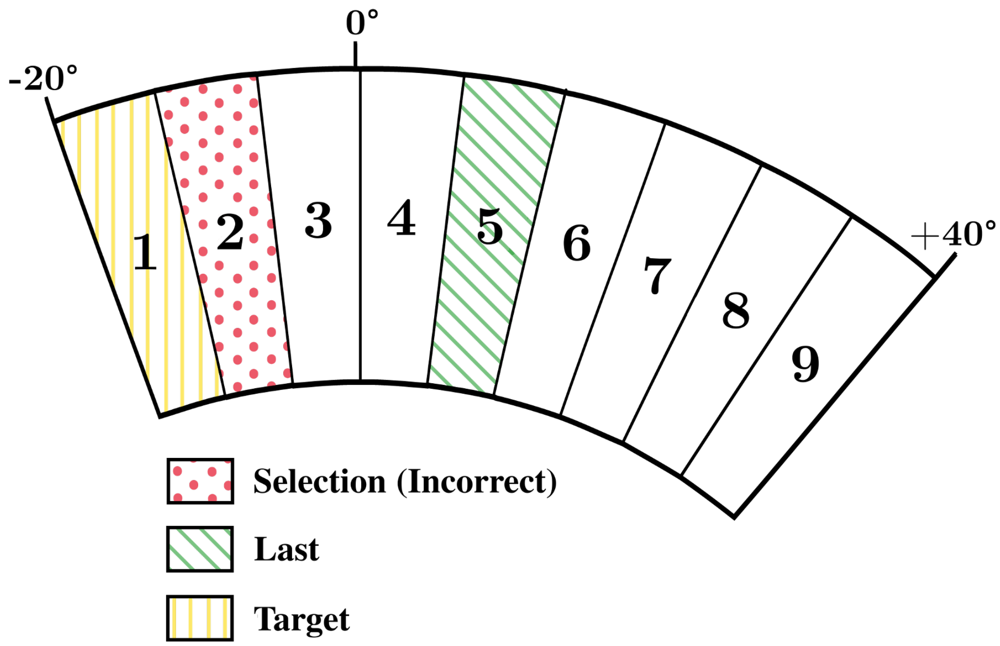
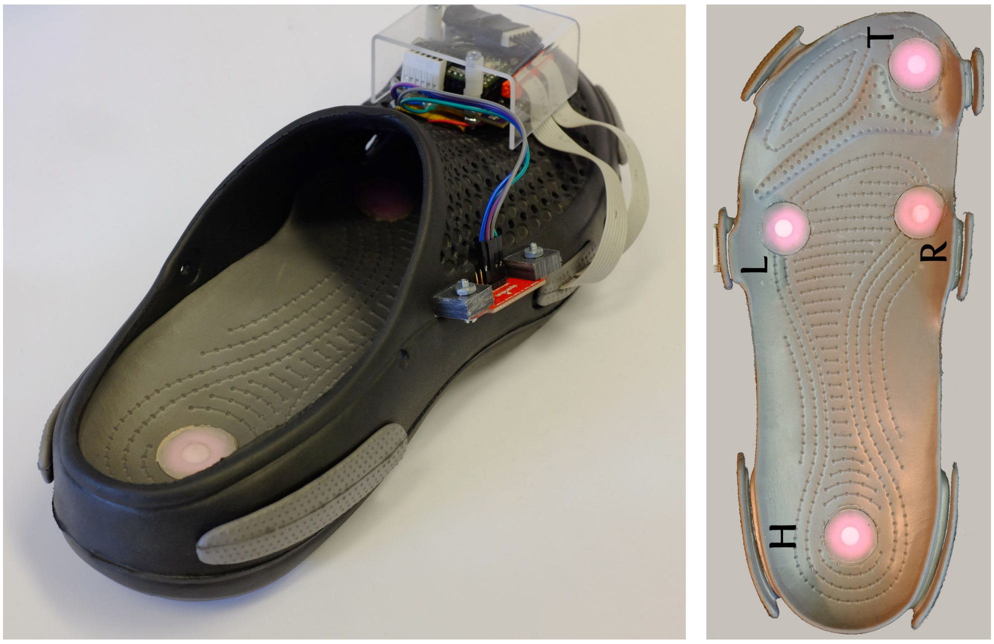


Figure C.6: Example menu for nine items. Starting from the last selection of item 5 (green diagonal hash), the participant has been instructed to select the target, item 1 (yellow vertical hash), but instead makes an incorrect selection of item 2 (red dots).

is least mentally demanding and offers the highest accuracy [14]. For these reasons, this foot rotational movement has been frequently used an input method. Recently, research on foot-based menu control have studied where range of foot rotation was separated into several areas and assigned either menu items [15] or different functions [13]. These findings motivated our evaluation of a 1-D menu selection task in order to improve our understanding of how haptic feedback should be delivered for maximum effectiveness. Specifically, we are interested in its usefulness for wearable devices and assessing the feasibility of using commodity sensor technology for such a system.

C.2.3 Methodology

(jan: *changed section title*) (jan: *Methods is more suitable to the actual content of the section and a commonly used heading for this. we also have subsection specifically for the xp design below.*)



(a) Shoe (b) Insole

Figure C.7: Haptic shoe and insole with actuator placement locations at **t**oe, **h**eel, **r**ight and **l**eft (first and fifth metatarsal head)

Participants had to choose a highlighted item from a radial menu by rotating their foot. In order to isolate foot interaction to item selection [3] and minimize the risk of parasitic foot movements, users confirmed their menu selections by pressing the space bar on a computer keyboard. Participants were only permitted to advance to the next trial after successfully selecting the indicated target item and could try as often as they needed.

In a pilot study, we found visual feedback on the current menu item to enable near-perfect accuracy. However, since we are interested in supporting users in tasks for which visual attention may be in high demand, we wanted to explore the feasibility of relying on haptic feedback as a substitute for the visual modality. Thus, we chose not to display current menu item during the experiment, but only displayed starting foot position, i.e., the last correct selection, and highlighted the chosen item after it was confirmed. (jan: *moved, rewritten*)

The menu was placed in the region of comfortable foot rotation, from -20° to 40° [15]. We compared three sizes of menus with three, six and nine items respectively,

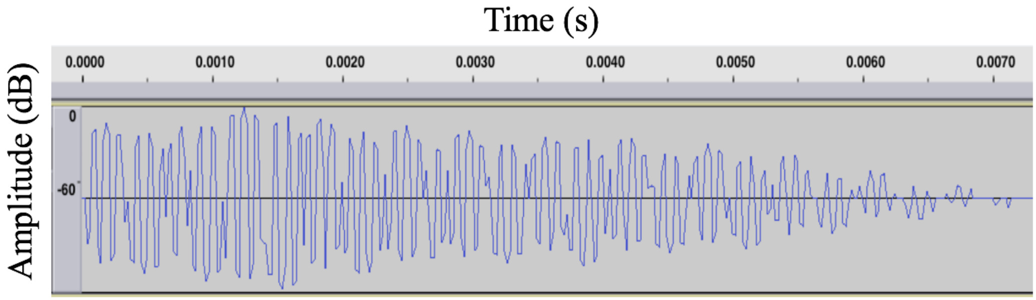


Figure C.8: Auditory feedback: click sound

dividing this range into equal slices. Thus, in menus with three items, the slices subtended 20° per item, 10° per slice for six items, and 6.7° per slice for a menu with nine items. The smallest of these angular displacements is close to the 5° increments we found to be a practical limit for accurate control of foot rotation in a parallel foot control study that relied on visual-proprioceptive feedback. (jan: *reworded previous paragraphs*) Fig. C.6 shows an example of a nine-item menu.

The menu interface was rendered graphically using the Unity game engine, running fullscreen on a desktop computer with a 24" flat panel display. Participants sat comfortably in front of the computer on a desk, with the foot under the table. Auditory feedback and pink noise to mask audible artifacts from the vibrotactile insole were provided using Beyerdynamic DT770 Pro M headphones.

Instrumented Haptic Shoe

Users controlled the menu through the TROG v2, worn on their right foot, pictured in Fig. C.7 (a). The integrated Bosch BNO-055 sensor provided us with a stable yaw reading through its integrated fusion of accelerometer, gyrometer and magnetometer readings (at 100 Hz). In order to minimize communication latency, menu item calculation was performed on the shoe's ARM Cortex-M4 microcontroller. Similarly, although the shoe supports wireless communication for future in-situ testing, we opted for a wired connection through USB to improve latency and reliability.

We used only the left (L) and right (R) actuator in this study, as illustrated in Fig. C.7 (b). shoe also contains four vibrotactile actuators.

C.2.4 Experiment Design

(jan: *reordered*) We evaluated 12 participants (8 male, 4 male) with an mean age of 24.6 years (SD = 4.4 years), with an average US men's shoe size of 9.9 (SD = 0.95), all reporting a dominant right foot. Participants gave informed consent and filled a pre- and post questionnaire. (jan: *pre-questions are reported, as are 1+2 post, 3rd did not yield anything interesting, so I consider this addressed.*)

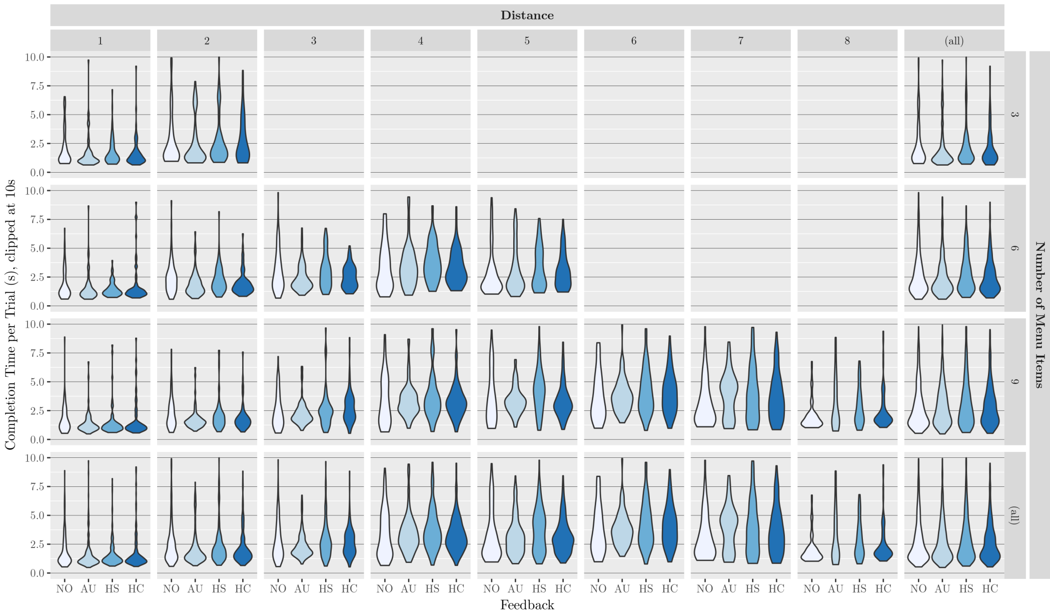


Figure C.9: Completion time: Violin plots with respect to distance, menu size and feedback condition

We compared four conditions for different feedback on the transitions between menu items:

- **NO**: No Feedback
- **AU**: Auditory Feedback
- **HC**: Haptic Click: Simultaneous click, left AND right
- **HS**: Haptic Side: Localized buzzing, left OR right

In the AU/HC/HS conditions, each step between menu items triggered a feedback event, either auditory or haptic. Auditory feedback (AU) consisted of a simple click sound [16], visualized in Figure C.8. Haptic click feedback (HC) was a simultaneous buzz of both left and right actuators simultaneously for 50 ms once for every menu item transition. Haptic side feedback (HS) was a localized buzz on the left or right actuator on the side the user rotated their foot towards. Both haptic feedback stimuli were approximately 50 ms in length, as a compromise between being short enough to allow for acceptable foot rotation speed without overlap and ability of pilot subjects to perceive them.

For the order of trials, we followed similar sequences as the multidirectional tapping task described in the ISO9241-9 standard (2002) [17] to cover all possible direction and distance combinations. An example sequence of targets for a three items menu would be 1, 3, 2, 3, 1. Participants carried out a series of ten randomly preselected trials in each of the four conditions, for a total of $10 \times 4 = 40$ trials, with conditions presented in the same order as in the test round.

After a training round for each menu size (3/6/9) and each condition (NO/AU/HC/HS), participants completed blocks of trials for each condition, with conditions ordered by Latin squares. Within each block, trials progressed from the smallest to largest menu size: 12 trials for the three-item menu, 30 trials for the six-item menu, and 48 trials for the nine-item menu, for a total of $4 \times (12 + 30 + 48) = 360$ trials per participant. Although learning effects may have helped participants achieve better performance as the experiment progressed, the results, described below, indicate that difficulty increased as a function of the number of menu items.

C.2.5 Results and Discussion

We compared completion time (time until a correct select was made) by menu size and feedback condition, and also analyzed the number of attempts with respect to distance and menu size according to feedback condition. In the three-item menu, participants achieved similar average completion time in all conditions. Fig. C.9 shows the completion times of participants as violin plots. These are similar to box plots but also illustrate the distribution of the data as the width of the "violin body". This allows us to see the multiple peaks corresponding to repeated attempts at hitting the correct target, e.g., for a distance of seven steps. (jan: *As mentioned by email: I learned this week that the* ***Friedman*** *test was apparently not appropriate to use.*) (jan: *leaving out: We used a Friedman test to investigate the difference in completion*

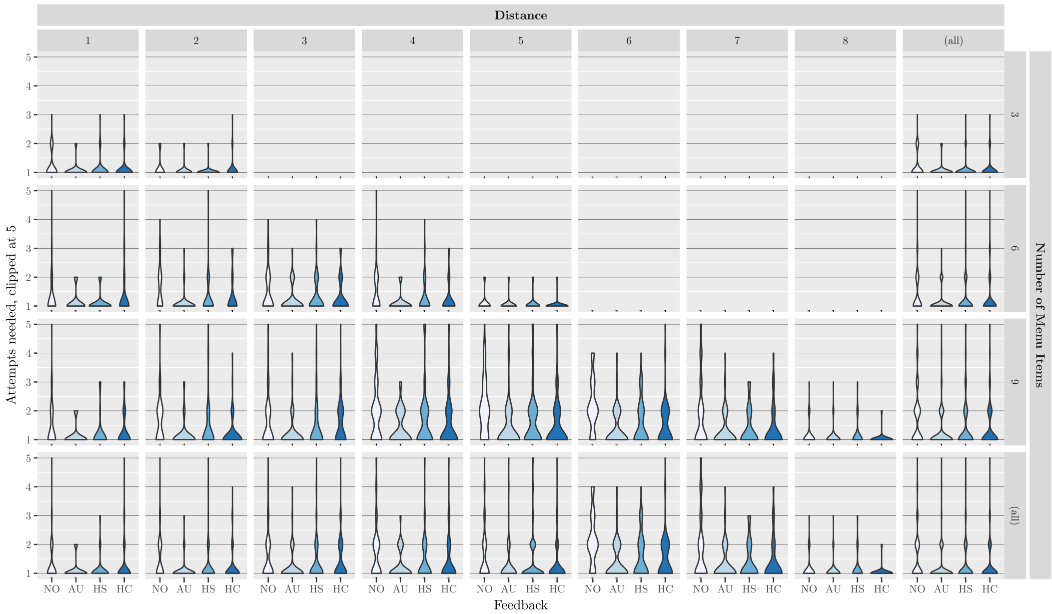


Figure C.10: Attempts needed: Violin plots with respect to distance, menu size and feedback condition

time between all experimental conditions. Following Bonferroni-Holm correction for multiple comparisons, no statistically significant differences were observed.)

The percentage of trials correct on the first attempt decreases with the number of menu items, and, as expected, is the worst for the no-feedback condition (see Table C.1). Interestingly, participants performed better in the clockwise direction. Given that all participants reported a dominant right foot, this may have anatomical reasons and should be investigated in future work.(jan: ***reworded**, could not find a suitable citation*) Required attempts, along with completion time, increased with increasing menu size.

As is evident from the plot of needed attempts in Fig. C.10,

moving to either the left or right extreme of the menu was performed easily, because of the infinite target width of an "edge" menu item. However, smaller step sizes proved to be more difficult, even with feedback. Movements within the larger menu required more attempts to reach the target item. 9 of 12 participants preferred audio feedback as modality, and all but one reported the system to be comfortable and easy to use.

Table C.1: Percentages of trials correct on first attempts, by condition, movement direction and menu size

Size	**AU**		**HC**		**HS**		**NO**	
	$\circlearrowright$	$\circlearrowleft$	$\circlearrowright$	$\circlearrowleft$	$\circlearrowright$	$\circlearrowleft$	$\circlearrowright$	$\circlearrowleft$
3	94.4	93.1	90.3	93.1	97.2	91.7	83.3	79.2
6	92.2	86.7	86.7	82.8	83.9	84.4	80	60.6
9	81.2	79.2	75	70.8	71.5	69.1	55.2	48.6

Audio feedback outperformed both haptic feedback conditions, and HC exhibited slightly better performance than HS as the number of items increased. However, the effect size was small. (jan: *leave out: , so we did not conduct a test for statistical significance.*)

C.2.6 Conclusion and Outlook

We investigated whether audio or haptic feedback would improve performance in menu selection when carried out by rotating the foot. Although audio was found to be the most effective modality, it may not be suitable for many potential applications, and the haptic alternative's drop in accuracy would not be meaningful in all applications. Localized haptic feedback did not show a benefit in our experiment, but may be helpful in real-world environments with relatively strong distractors or high loads on the somatosensory system. The system worked well despite employing only an inexpensive motion sensor for angle determination. We conclude that radial foot-menus with non-visual feedback on menu position show promise to be implemented effectively in a mobile context, which could enable many potential use cases of control input where the hands and eyes are occupied.(jan: *rewritten*) Future studies should both employ a distractor task, as well as haptic or audio noise to better represent real-world conditions.

Acknowledgments

We thank Pascal Fortin, Jeffrey Blum, André Arnold and Kathrin Krieger for their invaluable help.

C.3 Discussion and Outlook

Shoe 3

Modularize, Hifi Splashing [18], reproduceable? versions Foot shape varies greatly between individuals [19] and need to be accounted for [20].

C.3.1 Stochastic Resonance and Sub-Threshold Balance Aids

[21]

C.3.2 Energy Harvesting

Lastly, for a system that is to be used in everyday situations, energy consumption is a critical issue. As such, the possibility of energy harvesting from the large amounts of force involved in locomotion should be considered [22], [23].

Fluid Modulation

(ToDo: *choi et al:* **https://ieeexplore.ieee.org/abstract/document/8816165**)

(ToDo: *(Gait measure through fluid pressure [24])*) (ToDo: *Adaptive Shoe Soles [25] [26] etc*) (ToDo: *[27][28]*) (ToDo: *[29]*) (ToDo: *Used for foot fall dampening extensively: [30] [31]*) (ToDo: *VR Terrain Paper w/ Bladder Shoe [21],* **https://ieeexplore.ieee.org/abstract/document/6942715/**) (Add Citation: *for biofeedback in gait [32]*) Haptic perception through the feet informs a wide range of dynamic and static human activity. Stimulating the foot, for example to render virtual ground surface reactions, requires comparatively strong, and thus large actuators due to their placement between the ground and a human loading the foot. In stationary setups, actuators can be integrated into static assemblies at the ground surface. However, this is not feasible for mobile applications, for which the stimuli must be provided wherever the user happens to be. In such scenarios, delivery of sufficiently strong stimuli through conventional haptic actuators, such as voice coils, poses a significant challenge in terms of the associated electrical power requirements. Our invention instead leverages the foot-ground reaction forces available from standing or walking to help render the stimuli with sufficient force, thereby overcoming this challenge.

Taking inspiration from the design of Berrezag, Visell and Hayward (Compressibility and Crushability Reproduction through an Amorphous Haptic Interface, EuroHaptics 2012), the invention comprises the following elements.

1. one or more bladders made out of a strong sheet material, such as oriented polymers, embedded into footwear, and containing either air or a fluid, which may be either compressible or incompressible

2. the bladders are surrounded by or incorporate a flexible support structure, e.g., springs or foam, that decompress (i.e., expand if in a compressed state) the structure when it is not loaded (i.e., foot is in the air)

3. a valve or equivalent mechanism is used to regulate the flow in and out of each bladder chamber

4. if a fluid is used within the bladders, a reservoir chamber is also required, connected to each bladder via the valve; otherwise, if air is used, the bladders can refill from the surrounding air

5. pressure sensors are used to sense the loading of the footwear

When the user takes a step with such footwear, the sensors detect the loading and engage a controlled release of the fluid contained in one or more chambers. For an air-filled bladder, the valve allows air to bleed out of the bladder; otherwise, the fluid flows to a reservoir chamber. Liquids would offer greater fidelity since pressure waves propagate faster in liquid than in gas. Use of a magnetorheological fluid and a coil to modulate flow in the valve-mechanism would offer the advantage of avoidance of moving parts.

Control of the valve-mechanism during foot-ground contact allows for simulation of different textures, perceived by the user through the foot as haptic, and possibly acoustic, stimuli as the resulting pressure waves propagate through the entire system. DC components of the control signal would modify the compliance of the footwear, whereas non-DC components would correspond to higher-frequency haptic percepts, such as the various crumpling and impact effects associated with different ground surfaces, similar to those we have been able to simulate through various signal parameters in the "Floor-based haptic communication system" (patent 9041521).

When the user lifts the foot again, the flexible support structure refills the bladders.

The described architecture could also be used to implement variably compliant surfaces, either for simulation or destabilizing the user (e.g., in physical therapy).

Several years ago, our colleague, Vincent Hayward, described to us his team's approach to a fluid-chamber-based haptic rendering system that was subsequently published in EuroHaptics 2012 (http://dl.acm.org/citation.cfm?id=2352210). This work inspired us, as we were intrigued by the possibility of delivering perceptibly strong haptic stimuli to the feet without being restricted to our dedicated lab environment involving large tactile transducers. As our own research progressed to mobile architectures, we became further interested in exploring possibilities for rendering such haptic effects without the typical cost of high power demands. The possibility of harvesting some of the energy inherent in footsteps from everyday walking activity led to our realization that Hayward et al.'s approach could well be adapted for this purpose.

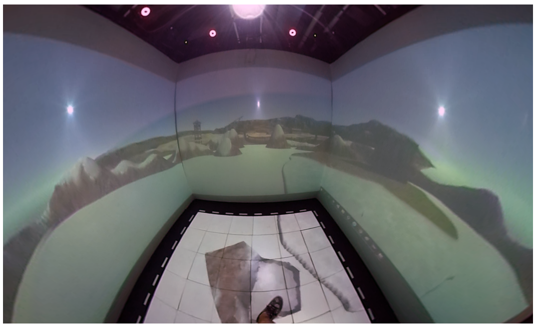
C.4 Conclusion

Comparatively little research exists on rendering vibrotactile patterns, or tactons, to the sole of the foot. However, it may be a natural placement location for vibrotactile stimulation for navigation tasks, posture and gait rehabilitation. Similarly, it may be more appropriate to render general haptic feedback to the foot when such vibrations would be distracting if delivered to other body parts. This motivated our development of a custom instrumented haptic insole, which incorporates four actuators, designed to deliver vibrotactile stimulation to the foot. As an initial test, we conducted a user study measuring recognition rates of tactons rendered during quiet stance and gait. Since actuator coupling and foot sensitivity change during gait, our insole uses a proximity sensor to detect footfall, allowing for consistent triggering at a defined point in the gait cycle when the actuators should be best coupled to the sole of the foot. We found that although overall recognition while standing (>92%) is much better than while walking (>66%), the majority of errors are confined to patterns that begin with the same actuator. Our initial evaluation shows promise for application of the insole, such as for sensorimotor rehabilitation and general human-computer interaction.

C.5 Acknowledgments

We thank Bobak Hamed-Baghi, Jeffrey Blum, Felix Dubé, Pascal Fortin, Daniel Horodniczy, Bei Chen Liu and Sebastian Zehe for their invaluable assistance throughout this project.

C.6 Publications and Disclosures

Application: CAVE to go: 
Foot-based exploration of Geospatial Data
(Hieda, Anlauff, Smith, Visell, Cooperstock, accepted at MMSP'16)

ROI McGill 17018: Anlauff, Cooperstock: In-shoe Fluid Modulation through Leveraged Ground-Reaction-Force, 2016 ROI McGill: Anlauff, Cooperstock: Posturally Informed Wearable Vibrotactile Interface ROI McGill Anlauff, Tordini, Cooperstock: Modular Haptic Actuator Inserts for Improved Inter-Actuator Decoupling and Mechanical Robustness

Inserts

In wearable haptic applications, vibrotactile actuators need to transfer their output energy to the user's skin. Misperception caused by cross-coupling of actuators in wearable haptics is more likely in systems in varying contact with the body, such as in our haptic insole system. We describe a method improving this coupling for use with multiple actuators in one system, while improving robustness. — During the development of an insole with multiple vibrotactile actuators, we discovered that there were two main challenges with regards to the embedding of the actuators. First, the vibrations generated by the actuators dissipate throughout the entire insole, making it hard to identify the active actuator among the set of actuators. Second, adequate mechanical protection of the actuators and their electrical connections from the forces involved in human gait is key to a robust system.

We developed modular inserts for the actuators that consist of a rigid, dished aluminum disk and a silicone cast filling in which the actuators are embedded. In our prototype, the pancake-type Linear Resonant Actuators (LRAs) are flush with the top surface of the insole, and thus in good contact with the user's foot, but mechanically decoupled from the rest of the shoe. This effect is enlarged by the dished aluminum disk that works as a baffle, akin to the housing of a loudspeaker.

The effect is that the user can perceive each actuator clearly located, and most of the energy from the actuators is transferred directly to the user's body, resulting in improved system efficiency. The compound used to embed the actuators (silicone, or other) can be chosen to optimize the impedance matching between the actuator and the user's body.

A benefit of this integration is that the actuators themselves are sealed into an elastic compound, which protects them and their wire connections, from environmental damage and excessive forces. This improved robustness is especially relevant for use in wearables such as smart insoles. The actuators are replaceable through the opening in the disk in case of damage.

What is novel?

Reduced actuator cross-coupling and improved robustness by providing a baffle for the mechanical energy produced by haptic actuators. What are the uses (applications)? Mobile delivery of haptic stimuli, for example on the feet. How is it non-obvious? In the wearable haptic systems we surveyed, the actuators are not embedded in inserts to improve the decoupling from each other while improving coupling with the body. This required a design combining soft and hard materials.

Tactile Communication System

We developed a system of multiple, co-located sensor-actuator combinations, for wearable haptic applications. This allows the system to provide vibrotactile stimulation directly to the site of the measurement.

This invention arose from our work on rendering the haptic textures of virtual materials that we have been able to simulate through various signal parameters in the "Floor-based haptic communication system" (ROI 09013). In order to render these inside a shoe, our initial implementation involves a stacked configuration of sensors and actuators, integrated into a smart insole. Our system can also be used as a general-purpose, foot-based human-computer interface, although we imagine that the techniques described here could be applicable to other body parts as well.

In our case, the actuators are vibrotactile and voice-coil based, such as linear resonant actuators (LRAs), TactileLabs Haptuators, or Auto-calibrated Haptic Actuator with Integrated Signal Amplification design (ROI 17035). Under each of the actuators, one force-sensing resistor measures the force loading of the insole region and the actuator. Globally, a 9-DoF motion sensor provides fused orientation headings and movement information. Our prototype platform also includes a sensor measuring proximity to the ground, enabling delivery of the vibrotactile stimuli at certain points in the gait cycle. A flex sensor measures gait-relevant shoe movement. The combination of these sensors allows us to capture many of the articulations of the foot, whether on the ground or in mid-air. In the future, accelerometers at the actuator sites could also measure actuator coupling (similar to the approach described by our previous publication [1].

The combinations of actuators and force sensors are placed in positions of articulated postural activity. In the case of the shoe, these are the regions where the foot exerts most pressure on the ground: big toe, first and fifth metatarsal head, and heel, and thus, improve the likelihood of coupling between the system and the user's body.. Furthermore, enabling such bi-directional communication between user and computer is highly relevant in situations where the hands are occupied. Additional benefits and capabilities of this approach include: • capture of postural expression, e.g., shift of the user's weight over their support base • interfaces that respond directly to gestures, e.g., tapping the big toe as a "button click" with simulated force-feedback • rendering of haptic effects with a higher degree of fidelity than systems with only a global sensor-actuator channel • rendering of rich tactile notifications, by incorporating multiple sensor-actuator combinations and exploiting sensory illusions, such as cutaneous saltation or haptic funnelling (cf. Disney's "Tactile Brush" algorithm)

As opposed to a conventional switch, the system can take into account different force profiles to adapt to the user, or create a feedback loop that links actuator activation to readings from the sensor, similar to Apple's Taptic engine for finger-based inputs, either simulating a physical micro-switch or providing richer feedback, e.g., during dragging operations.

An example application could be personalized delivery of haptic alerts to specific staff members of a hospital operating room, in which the staff would benefit not only from the reduced attentional load, but would have constant, immediate access to a control mechanism to confirm receipt of the alarm, silence it, or perform other operations, with haptic feedback providing confirmation of the selected action. In another application, feedback for gait physiotherapy, the invention could support optimally timed delivery of the haptic stimuli, capitalizing on the measurements from the proximity sensor to deliver feedback just before footfall, when human tactile sensitivity is highest.

1. Blum, J., Frissen, I., and Cooperstock, J. R., Improving Haptic Feedback on Wearable Devices through Accelerometer Measurements. User Interface Software and Technology (UIST), November 2015. ACM. What is novel?

Combining sensors and actuators in posturally salient points for wearable multichannel haptic feedback. What are the uses (applications)? Virtual Reality, wearable alarm/communication and control systems. How is it non-obvious? To the best of our knowledge, existing haptic systems did not co-locate sensors and actuators, and did not position them in anatomically relevant locations.

References for Chapter C

[1] S. " Lee and T. Starner, "BuzzWear: Alert Perception in Wearable Tactile Displays on the Wrist," in *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems (CHI '10)*, ser. CHI, New York, NY, USA: ACM, 2010, pp. 433–442, ISBN: 978-1-60558-929-9. DOI: 10.1145/1753326.1753392. [Online]. Available: http://doi.acm.org.proxy2.library.mcgill.ca/10.1145/1753326.1753392.

[2] M. Azadi and L. Jones, "Evaluating vibrotactile dimensions for the design of tactons.," *IEEE transactions on haptics*, vol. 7, no. 1, pp. 14–23, 2014. [Online]. Available: http://europepmc.org/abstract/MED/24845742.

[3] E. Velloso, D. Schmidt, J. Alexander, H. Gellersen, and A. Bulling, "The Feet in Human–Computer Interaction: A Survey of Foot-Based Interaction," *ACM Comput. Surv.*, vol. 48, no. 2, 21:1–21:35, Sep. 2015, ISSN: 0360-0300. DOI: 10.1145/2816455. [Online]. Available: http://doi.acm.org/10.1145/2816455.

[4] A. Pedotti, R. Assente, and G. Fusi, "Multisensor piezoelectric polymer insole for pedobarography," *Ferroelectrics*, vol. 60, no. March, pp. 163–174, 1984, ISSN: 0015-0193. DOI: 10.1080/00150198408017519.

[5] J. A. Paradiso, K. Hsiao, A. Y. Benbasat, and Z. Teegarden, "Design and implementation of expressive footwear," *IBM Systems Journal*, vol. 39, no. 3.4, pp. 511–529, 2000, ISSN: 0018-8670. DOI: 10.1147/sj.393.0511. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=5387021.

[6] D. Zanotto, L. Turchet, E. M. Boggs, and S. K. Agrawal, "SoleSound: Towards a novel portable system for audio-tactile underfoot feedback," *5th IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics*, pp. 193–198, 2014. DOI: 10.1109/BIOROB.2014.6913775. [Online]. Available: http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=6913775.

[7] A. Law, J. Ip, B. Peck, Y. Visell, P. Kry, and J. R. Cooperstock, "Multimodal floor for immersive environments," in *SIGGRAPH Emerging Technologies*, New Orleans, LA, USA: ACM, Aug. 2009, 16:1. DOI: 10.1145/1597956.1597972.

[8] A. A. Priplata, B. L. Patritti, J. B. Niemi, R. Hughes, D. C. Gravelle, L. A. Lipsitz, A. Veves, J. Stein, P. Bonato, and J. J. Collins, "Noise-enhanced balance control in patients with diabetes and patients with stroke.," *Annals of Neurology*, vol. 59, no. 1, pp. 4–12, Jan. 2006, ISSN: 0364-5134. DOI: 10.1002/ana.20670. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/16287079.

[9] J. Watanabe and H. Ando, "Pace-sync shoes: Intuitive walking-pace guidance based on cyclic vibro-tactile stimulation for the foot," *Virtual Reality*, vol. 14, pp. 213–219, 2010, ISSN: 13594338. DOI: 10.1007/s10055-009-0137-y.

[10] A. Meier, D. J. C. Matthies, B. Urban, and R. Wettach, "Exploring Vibrotactile Feedback on the Body and Foot for the Purpose of Pedestrian Navigation," in *Proceedings of the 2nd International Workshop on Sensor-based Activity Recognition and Interaction*, ser. WOAR '15, New York, NY, USA: ACM, 2015, 11:1–11:11, ISBN: 978-1-4503-3454-9. DOI: 10.1145/2790044.2790051. [Online]. Available: http://doi.acm.org/10.1145/2790044.2790051.

[11] I. Yoda, K. Ito, and T. Nakayama, "Modular Gesture Interface for People with Severe Motor Dysfunction: Foot Recognition.," *Studies in health technology and informatics*, vol. 242, p. 725, 2017.

[12] J. Scott, D. Dearman, K. Yatani, and K. N. Truong, "Sensing foot gestures from the pocket," in *Proceedings of the 23nd annual ACM symposium on User interface software and technology*, ACM, 2010, pp. 199–208.

[13] B. Hatscher, M. Luz, and C. Hansen, "Foot Interaction Concepts to Support Radiological Interventions," vol. 17, no. September, pp. 3–13, 2017. [Online]. Available: http://dx.doi.org/https://doi.org/10.18420/muc2017-mci-0174.

[14] C. Grane and P. Bengtsson, “Menu Selection with a Rotary Device Founded on Haptic and/or Graphic Information,” *Proceedings - 1st Joint Eurohaptics Conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems; World Haptics Conference, WHC 2005*, pp. 475–476, 2005. DOI: 10.1109/WHC.2005.90.

[15] K. Zhong, F. Tian, and H. Wang, “Foot menu: Using heel rotation information for menu selection,” in *Proceedings - International Symposium on Wearable Computers, ISWC*, 2011, pp. 115–116, ISBN: 9780769544380. DOI: 10.1109/ISWC.2011.10.

[16] freesound.org User cabled_mess, *Minimal UI Sounds: Click 01*, https://freesound.org/people/cabled_mess/sounds/370962/, 2018. [Online]. Available: https://freesound.org/people/cabled_mess/sounds/370962/ (visited on 01/10/2018).

[17] R. W. Soukoreff and I. S. MacKenzie, “Towards a standard for pointing device evaluation, perspectives on 27 years of Fitts’ law research in HCI,” *International journal of human-computer studies*, vol. 61, no. 6, pp. 751–789, 2004.

[18] G. Cirio, M. Marchal, A. Lécuyer, and J. R. Cooperstock, “Vibrotactile Rendering of Splashing Fluids,” *Transactions on Haptics*, vol. 6, no. 1, pp. 117–122, 2013, ISSN: 2329-4051. DOI: 10.1109/TOH.2012.34. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6226398.

[19] I. Krauss, S. Grau, M. Mauch, C. Maiwald, and T. Horstmann, “Sex-related differences in foot shape.,” *Ergonomics*, vol. 51, no. February 2015, pp. 1693–1709, 2008, ISSN: 0014-0139. DOI: 10.1080/00140130802376026.

[20] K. J. Mickle, B. J. Munro, S. R. Lord, H. B. Menz, and J. R. Steele, “Foot shape of older people: implications for shoe design,” *Footwear Science*, vol. 2, no. February 2015, pp. 131–139, 2010, ISSN: 1942-4280. DOI: 10.1080/19424280.2010.487053.

[21] Y. Wang and M. A. Minor, "Design of a bladder based elastomeric Smart Shoe for haptic terrain display," *IEEE International Conference on Intelligent Robots and Systems*, no. Iros, pp. 1236–1241, 2014, ISSN: 21530866. DOI: 10.1109/IROS.2014.6942715.

[22] T. G. McKay, B. M. O'Brien, and I. A. Anderson, "A Technology for Soft and Wearable Generators," *Smart Actuation and Sensing Systems-Recent Advances and Future Challenges, G. Berselli, R. Vertechy and G. Vassura, Eds., InTech publishers*, p. 716, 2012.

[23] K. Ylli, D. Hoffmann, B. Folkmer, and Y. Manoli, "Design, fabrication and characterization of an inductive human motion energy harvester for application in shoes," *Journal of Physics: Conference Series*, vol. 476, p. 012 012, Dec. 2013, ISSN: 1742-6588. DOI: 10.1088/1742-6596/476/1/012012. [Online]. Available: http://stacks.iop.org/1742-6596/476/i=1/a=012012?key=crossref.7c8458c3c2ca7c4d61cb04860c7a9564.

[24] K. Kong and M. Tomizuka, "A gait monitoring system based on air pressure sensors embedded in a shoe," *Mechatronics, IEEE/ASME Transactions on*, vol. 14, no. 3, pp. 358–370, 2009. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=4813239.

[25] F. E. Ellis, *Footwear soles including an electronic control system controllining the flow resistance of a magnetorheological fluid in compartments with internal flexibility sipes between inner and outer compartments*, 2014. [Online]. Available: https://www.google.com/patents/US20140331523.

[26] ——, *Microprocessor control of magnetorheological liquid in footwear with bladders and internal flexibility sipes*, 2016. [Online]. Available: https://www.google.com/patents/US9271538.

[27] T. Kikuchi, Y. Masuda, T. Mitsumata, and S. Ohori, "Basic Study on Intelligently Haptic Controllable Shoes with Magnetic-Field Sensitive Gel," in *Proceedings of the 20th International Conference on Artificial Reality and Telexistence (ICAT2010)*, Adelaide, Australia, 2010, pp. 160–

165, ISBN: 9784904490037. [Online]. Available: http://www.ic-at.org/ICAT2010_Proceedings/poster/paper-101.pdf.

[28] T. Kikuchi, Y. Masuda, M. Sugiyama, T. Mitsumata, and S. Ohori, "Prototype of haptic device for sole of foot using magnetic field sensitive elastomer," *Journal of Physics: Conference Series*, vol. 412, p. 012 036, Feb. 2013, ISSN: 1742-6588. DOI: 10.1088/1742-6596/412/1/012036. [Online]. Available: http://stacks.iop.org/1742-6596/412/i=1/a=012036?key=crossref.39c02eb8439104d9b88ddac6723c4d67.

[29] A. Berrezag, Y. Visell, and V. Hayward, "Compressibility and crushability reproduction through an amorphous haptic interface," in *Proceedings of the Eurohaptics Symposium*, 2012, pp. 186–190. [Online]. Available: http://link.springer.com/chapter/10.1007/978-3-642-31404-9_32.

[30] M. F. Rudy, *Insole construction for articles of footwear*, 1980. [Online]. Available: https://www.google.com/patents/US4183156.

[31] P. W. Auger, D. R. Potter, and B. J. Kilgore, *Shoe bladder system*, 1993. [Online]. Available: http://www.google.de/patents/US5257470.

[32] Z. McKinney, K. Heberer, B. N. Nowroozi, M. Greenberg, E. Fowler, and W. Grundfest, "Pilot evaluation of wearable tactile biofeedback system for gait rehabilitation in peripheral neuropathy," in *Haptics Symposium (HAPTICS), 2014 IEEE*, Feb. 2014, pp. 135–140, ISBN: 9781479931316. DOI: 10.1109/HAPTICS.2014.6775445. [Online]. Available: http://ieeexplore.ieee.org/xpls/abs_all.jsp?arnumber=6775445.

D Further Applications of the Modular Toolkit

(I 90%, S 80%) (
words)

D.1 Brain-Computer Interfaces for Minimally Responsive Patients

Context

Population: minimally responsive patients

- Anually several thousand severe traumatic brain injury cases, with often permanent partial loss of consciousness and communication ability (citation needed), e.g., through accidents or stroke.
- The disorders of consciousness (DOC) can be classified based on arousal and awareness, Blume et al. [1] give an overview; diagnosis and distinction between vegetative state / unresponsive wakefulness syndrome (UWS), minimally conscious state (MCS), and locked-in syndrome is difficult.
- miss-diagnosis rates of 40% are high, with dramatic consequences for locked-in patients that would mentally able to communicate but which is often not recognized; limited means for therapy so far
- Amyotrophic lateral sclerosis (ALS), a progressive neurodegenerative disease causes the death of neurons controlling voluntary muscles can lead to a similar situation [2] (ToDo: *citation is about care*).

(ToDo: *P300 review* https://www.frontiersin.org/articles/10.3389/fneng.2012.00014/full)

- Use senses of touch and hearing together with event-related neural responses (measured through EEG) or residual motor-function to re-establish basic communication, e.g, to allow yes/no answers, or operation of a speller for text input.
- Existing approach: Mind Beagle [3] demonstrate the basic principle of P300 classification for BCI communication of DOC-affected individuals, but lacks training, is not suitable for long-term use (e.g., EEG-electrode grids), and don't offer training.
- Our Approach: communication system AND trainer for BCI-control as well as improved motor control of residual abilities

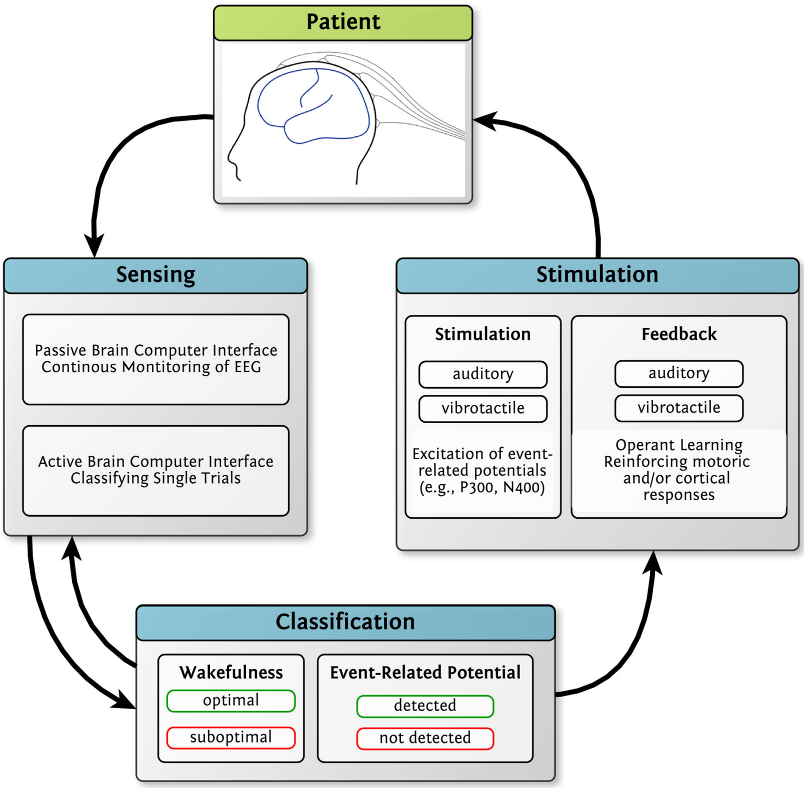


Figure D.1: NeuroCommTrainer: Brain-Computer Interface combining passive, long-term monitoring with interactive training. Figure courtesy of Andrea Finke.

- real-time signal analysis / classificator that adapts to the individuals brain responses [4]
- adaptated to the individual: choice of body parts for sensors/actuators, adaptive machine learning system
- Operation principle:
  - initially: establish that users can understand semantic speech content: measure N400 EEG response to sentences that don't make semantic sense
  - detect periods of optimal responsiveness through continous long-term monitoring of brain activity with novel electrode grids
  - inside such periods: either

        * offer individualized training for body parts (to eventually control an interface, such as a speller); example: spontaneous finger movements happen, so vibrotactile stimulate and train user to consciously control the finger more (voice: “this is your finger”), with positive feedback, or
        * for users that have to much spasticity, use P300 response to rare stimuli [5] to implement yes/no answers based of conscious focus on a body part (auditory and haptics tested, depending on patient ability; in current trial only haptic P300)
    - end interventions them when there are signs of fatigue
    - based on operant conditioning, always give positive rewards: familiar voice encouraging successful training / interaction with BCI

- real-world system, multi-disciplinary research cooperation (Ambient Intelligence, Neuroinfomatics for BCI, Psychology for study paradigms and execution, Neuropsychology for novel EEG/BCI systems, Health Science Ethics, with industry partners (improve long-term eeg comfort, novel electrodes, portable eegs; interfaces for operation by clinical staff) in collaboration with house ELIM Bethel where the studies take place.
- Our part here is the non-EEG/EMG hardware, that measures user motion and environment variables, such as sound levels; and delivers vibrotactile stimuli.

D.1.1 Requirements / Goals

Requirements:

- This requires robust and comfortable sensor and actuator wearables for in-situ 24/7 use.
- Each patient has different abilities and as such a modular system is needed to accomodate appropriate sensing and actuating modalities to each case, to creating an effective multi-modal brain-computer interface that that complies with clinical requirements.

Goals:

Non-Goals:

Constraints:

Evaluation: Provide a framework for assessing the quality of the system design decisions, Are the design decisions made by the authors the reason why the system achieved the goals?

Key Insights *idea about the **structure** of a problem*

Key Design Decisions *Rationale, provide wisdom to future implementors. E.g., "one common property of a preferred solution is that it is a simple one": eschew complexity that is not fundamental to meeting goals.*

Separate from implementation details, motivated by end-to-end view of the problem.

What were the alternatives considered at various points, and why were the choices made the way they were? Which side-streets did you not happen to explore?

D.1.2 System Architecture

The measurement modalities are shared between both modes, the trainig mode adds feedback. For brain activity, we recorded electro-encephalograpy (EEG) measured through a novel type of adhesive electrode grid worn on the forehead, feeding into a portable, wireless amplifier system by Smarting (channel count?). We measured motion of the user with a second smarting unit measured EMG responses on the arms (?), and up to four of the BNO055 sensors described in Section 4.2.5. A breathing sensor can be used to measure breathing rate, which is linked to arousal. We measure the ambience with the AmbiSense board (ToDo: *ref*), we measure ambient temperature, light level and color, and background sound levels with a 128-bin FFT.

In the long-term monitoring, we use these sensors to identify time periods of highest patient communication ability and wakefulness using a combination of sensors, adapted to the abilities of the individual.

In the short-term training, we detected event-related potentials (ERPs) through EEG, such as P300 responses, and rudimentary motor actions through the wearable sensors and EMG, e.g. trying to move a limb. P300 ERPs are elicited in decision processes in relation to two stimuli provided by the oddball-paradigm, with one being frequently presented, the other one rare. We can use it to create speller systems, allowing binary control input over a text grid (for example), by focussing on the letter they want to spell. The stimuli can be visual, auditory or haptic; here they are vibrotactile. (ToDo: *cite something* ***https://onlinelibrary.wiley.com/doi/full/10.1111/j.1469-8986.2006.00456.x***) In the training, we also provide stimuli to the user for example the voice of a relative instructing the patient ("I will touch your left hand") or familiar music and sounds (ToDo: *still unclear why we do this?*) in combination with gentle vibrotactile feedback to support the user in learning to control certain extremities better, that then can be used for control input.

Sensors and Tactors on Limbs: NCT Master

Example of modular system using a SBC as system master:

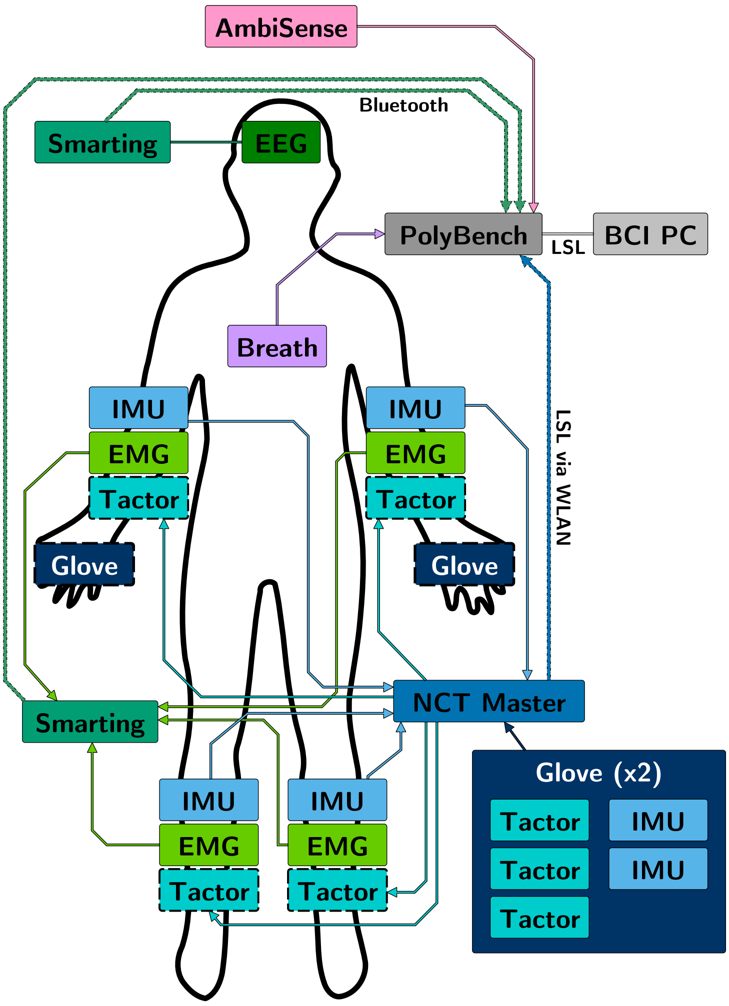


Figure D.2: NeuroCommTrainer System Overview. All connections are wired, except otherwise dashed lines. Dashed borders around components indicate that they are used only for the interactive mode.

- Combination of externally produced, off-the-shelf hardware, industry prototypes and our system
- Sensors:
  - Electrophysiologicals EEG / EMG through Smarting amplifiers that communicate with PolyBench computer through Bluetooth. Two units.
  - BNO055

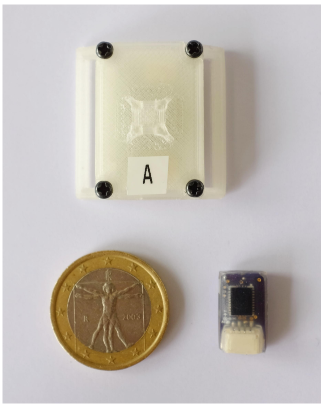


Figure D.3: BNO055 sensors, large version used on 1 inch straps, small version integrated in glove fingertips, used to detect inhabitant motion.

  - AmbiSense: Light color / intensity, humidity, temperature, tvoc, co2, 127 band FFT every n ms. apparently correlates well for light color. uart-lsl-bridge.py
  - Breathing rate sensor (details)
- Actuators: Lofelt on extremities, LRAs on hands.
- Communication via LSL over Ethernet/WLAN
- Consolidation, visualization and recording of data as well as control through a graphical interface through Applied Biosignals PolyBench.
- BCI system implemented in Python, running on "BCI Computer"
- Our contribution: "NCT Master" central controller that samples motion sensors and drives haptic actuators, Gloves (3x LRA, 2x BNO), BNO Straps, AmbiSense.

- Interconnect: Qwiic/JST SH for sensors, actuators and glove. Some external components via Bluetooth.
- NCT Master: Raspberry Pi 3 A+ combined with Audio-400 for haptic stimulus synthesis. Shown in Fig. D.4.

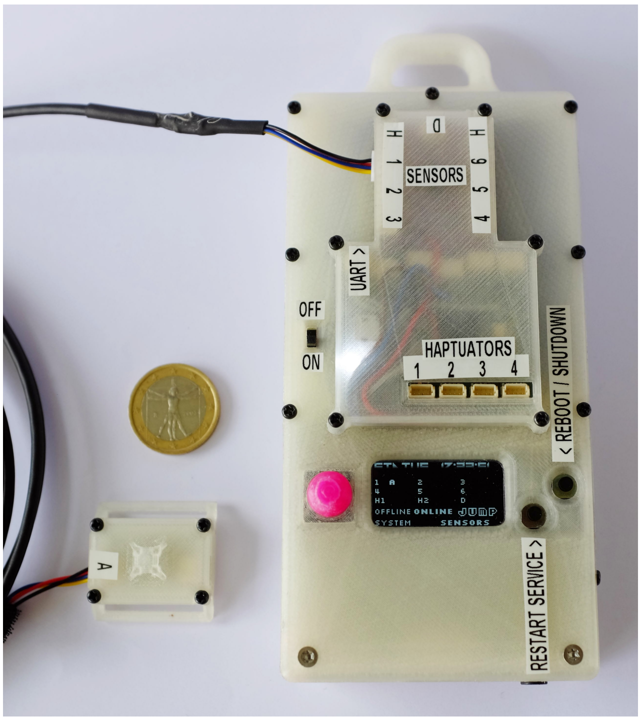


Figure D.4: NCT master with one BNO sensors connected

  - Power: Goal: run >8h autonomously. 8400 mAh battery + power management (lipo charger, 1.5A 3V3 regulator, battery gauge read by RasPi over I2C to allow charge status monitoring and automatic shutdown). System power consumption varies depending on number of sensors connected (1 BNO equals roughly 20 mA), but system runs well over 12h. Lithium polymer 18650 cells chosen for higher durability and less explosion risk.
  - 8 I2C/Qwiic ports, 6 arbitrated through I2C bus multiplexer to allow connecting up to 12 BNOs and two gloves (w/ 2 BNOs each). I2C is implemented through software-I2C, as the Raspi HW I2C does not support clock-stretching which the BNO requires. Cables up to 3m work well (shielded cables using Sensorcord).

- OLED display: display of auto-detected sensors and their motion for testing, battery and system status, restart of service/unit, shutdown. Running over the HW I2C interface.
- Communication through WLAN and LSL: for P300 detection, jitter is a big problem. LSL provides good timestamping and reliable reconnection mechanism for a robust real-world system. BLE was considered, but dropped after initial tests; throughput was not great, we would have needed to implement all the reconnection magic ourselves, and possibly had more problems with interference in the 2.4 GHz band than with 5 GHz wifi. Communication over a central unit close to the patient means more cables, but also less system complexity and latency.
- NCT master software: two main python services: one handling the user interface and system status streaming; one handling the readout and streaming of the sensors. Autodetects BNOs and creates LSL streams accordingly, handles reconnects. We stream gravity compensated acceleration and angular velocity at 80 Hz for 4 BNOs; limiting factor is the software I2C implementation running at 400 Khz. Battery status and sensor status are also streamed over LSL for remote supervision. GPIO handler for shutdown/reboot requests

D.1.3 Sensors and Tactors on Hands: Vibrotactile Glove

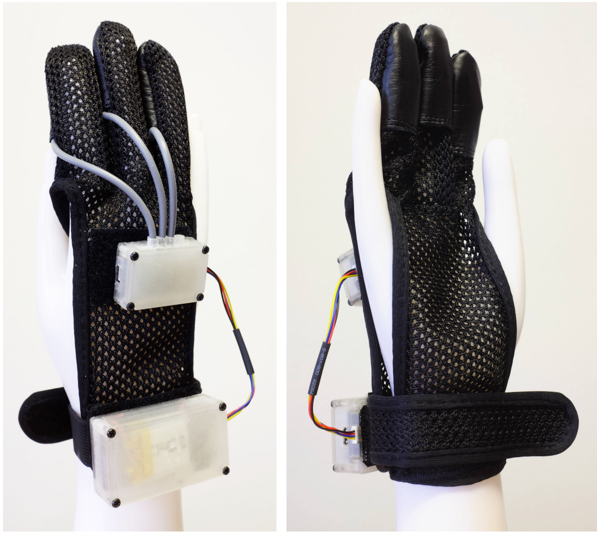


Figure D.5: Vibrotactile glove for stimulation of fingertips

Context

- P300 discrimination tests using the oddball paradigm.
- following successful pilot study with four coin cell ERMs driven from $BRIX_2$ with modified motor extension, taped to fingertips
- later application: body scheme training (ToDo: *wording*)

Requirements

Goals

- vibrotactile stimulation on fingertips, for P300 event-related potentials (ERPs) and eventual body scheme training
- measure motion of hand
- extensibility for later addition of motion sensors on fingertips

- two units, left and right hand
- garment should be easy to don and remove by care personel on hands of patients with possible spasticity
- improved robustness: pilot testing showed that hookup wires of actuators are very thin (maybe 30 AWG?), can rip easily and solder joints on actuators break easily, can not be easily replaced in the field
- one size fits most (fingers of minimally responsive patients are similar in diameter due to muscle atrophy)
- support wired and wireless / untethered operation
- support both standalone operation and in combination with NCT master described in previous section

NON-GOALS ??

CONSTRAINTS

- for ERPs, jitter is not acceptable
- space on fingertips is very limited
- repair not possible in the field

EVALUATION / HOW TO ASSESS THE QUALITY OF DESIGN DECISIONS?

- did the system mechanically survive the study?
- did the stimuli evoke reliable, large magnitude ERPs

KEY INSIGHT

- a glove with just fingertips connected to a wrist strap is sufficient and minimized complications
- not a full glove; three fingers are enough
- fail gracefully (plugs for interconnect)

Key Design Decisions

- use archery glove (index, middle, ring fingers)
- place electronics divided into two units on back of hand and on wrist
- fully encapsulate actuator wires
- use modular architecture

Alternatives Considered BRIX$_2$ prototype with non-encapsulated actuators

Implementation Details

- Controller: Adafruit Bluefruit nRF52 Feather, control / communication through USB Serial or Bluetooth Low Energy
- 150 mAh Lithium Polymer battery (runtime?)
- Feather wing with Qwiic interconnect for modular extension
- Bosch BNO055 Qwiic motion sensor extension: 3 axis accelero-/gyro-/magnetometer w/ sensor fusion for stable orientation, placed on back of hand
- QuadDRV Qwiic extension with 4 channel haptics drivers, placed on back of hand
  - 3 Channels used, one for each finger
  - Tactors: Jin Long LVM061530B rectangular Linear Resonant Actuators (LRAs)[1]
  - Drive/Resonancy Frequenz: 200 ±10 Hertz
  - Rare: Amplitude 16%, Duration 300 ms
  - Frequent: Amplitude 100 %, Duration 500 ms
- 3D-printed housings for Controller + Battery, tactor drivers + motion sensor, and tactors (glued into fingertips of gloves)
- Tubular wire housing connecting housing on both sides of the actuator, providing physical protection

[1] http://www.vibration-motor.com/products/download/LRA-LVM061530.pdf

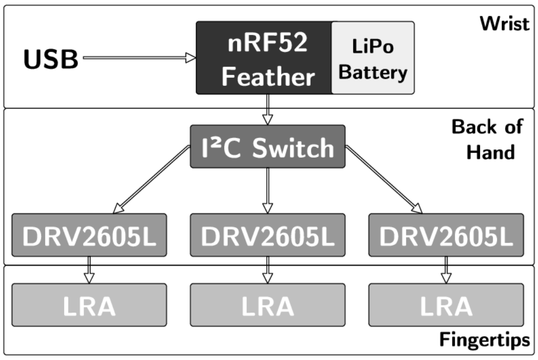


Figure D.6: Architecture of the vibrotactile glove.

Evaluation

- Are the design decisions made by the authors the reason why the system achieved the goals?
- Evidence?
- Does the system introduce new capabilities to the field?

D.2 Neumann-Cosel System

"Steady state somatosensory evoked potential Experiment"

Ingrediens:
- 6 Channels Lofelt
- Audio Adaptors w/ Power infusion, 8 Ch Logilink USB Audio Interface
- A humanoid robot

D.3 PhysioSon/Ago

The PhysioSon systemm supports a patient in absence of a therapist in the physiotherapy excercises with auditory feedback. Therapists can record "recipes" of a movement, for example squats, on himself, and the patient is guided by feedback on error (ToDo: *reference this to background*).

Squats, as an example, should be executed without extending the knees extend over the feet. Primary feedback parameter here is the relative angle difference between two MARG sensors, attached with elastic straps on the user's legs: one base controller (Section 4.2.1) above the knee, and one BNO055 sensor (Section 4.2.5) below. A controller application computer computes the difference of the euler angles of the two sensors. During the recording of the quat recipe, the therapist executes the motion repeatedly, and the range of this feedback parameter is stored. When the patient executes the recipe during excercise, the system provides feedback on a possible overextension through a sonification based on the overall sum of angle differences. The controller is written in Python and SuperCollider with the PyA binding, running on a desktop computer.[2]

With regard to our modular infrastructure, we can view this application as the prototypical engineering student project. Here, the goal was to understand the challenges for providing the most basic feedback. Feedback on knee flexion was part of the first exploratory prototype, and deriving such feedback without the use of an external motion capture system is a key component in biofeedback systems for motor learning. It provides us with a test scenario to gain understanding about sensor drift in real-world application.

[2] https://github.com/interactive-sonification/pya

D.4 Discussion

D.5 Publications

References for Chapter D

[1] C. Blume, R. del Giudice, M. Wislowska, J. Lechinger, and M. Schabus, "Across the consciousness continuum - from unresponsive wakefulness to sleep," *Frontiers in Human Neuroscience*, vol. 9, no. March, pp. 1–14, 2015, ISSN: 1662-5161. DOI: 10.3389/fnhum.2015.00105. [Online]. Available: http://www.frontiersin.org/Human_Neuroscience/10.3389/fnhum.2015.00105/abstract.

[2] E. V. Hobson and C. J. McDermott, "Supportive and symptomatic management of amyotrophic lateral sclerosis," *Nature Reviews Neurology*, vol. 12, no. 9, pp. 526–538, 2016, ISSN: 17594766. DOI: 10.1038/nrneurol.2016.111. [Online]. Available: http://dx.doi.org/10.1038/nrneurol.2016.111.

[3] C. Guger, B. Allison, A. Kammerhofer, F. Guttmann, T. J. Von Oertzen, R. Spataro, V. La Bella, J. Annen, S. Laureys, A. Heilinger, R. Ortner, and W. Cho, "MindBEAGLE - A new system for the assessment and communication with patients with disorders of consciousness and complete locked-in syndrom," *2017 IEEE International Conference on Systems, Man, and Cybernetics, SMC 2017*, vol. 2017-Janua, pp. 3008–3013, 2017. DOI: 10.1109/SMC.2017.8123086.

[4] A. Finke, I. Steppacher, J. Kissler, and H. Ritter, "Frequency Band Variations Predict EEG Single-Trial Classification Performance in Disorder of Consciousness Patients," *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, vol. 2018-July, pp. 1927–1930, 2018, ISSN: 1557170X. DOI: 10.1109/EMBC.2018.8512719.

[5] C. C. Duncan-Johnson and E. Donchin, *On Quantifying Surprise: The Variation of Event-Related Potentials With Subjective Probability*, 1977. DOI: 10.1111/j.1469-8986.1977.tb01312.x.

endsinglespace

Appendix

Checklist for good systems papers:

- Are the ideas in the paper new?
- How do you know?
- Can you state the new idea concisely?
- What exactly is the problem being solved? Be specific. Why couldn't it be solved just as well by existing techniques?
- Are the ideas significant enough to justify a paper?
- Is the work described significantly different from existing related work?
- Is all related work referenced, and have you actually read the cited material?
- Are comparisons with previous work clear and explicit?
- Does the work comprise a significant extension, validation, or repudiation of earlier but unproven ideas? Implementations are expensive!
- What is the oldest paper you referenced? The newest? Remember that citations not only acknowledge a debt to others, but also serve as an abbreviation mechanism to spare your reader a complete development from first principles.
- Does the paper describe a system that has actually been implemented?
- What have you learned from the work?
- What should the reader learn from the paper?
- How generally applicable are these lessons?
- What were the alternatives considered at various points, and why were the choices made the way they were? Which side-streets did you not happen to explore?
- Did the choices turn out to be right, and, if so, was it for the reasons that motivated them in the first place?
- What are the assumptions on which the work is based?
- Are they realistic?

- How sensitive is the work to perturbations of these assumptions?
- Does the introductory material contain excess baggage not needed for your main development?
- Do you include just enough material from previously published works to enable your reader to follow you thread of argument?