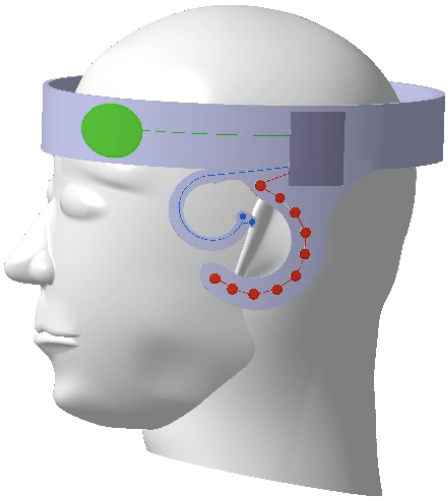
École Polytechnique Fédérale de Lausanne

A neural interface to facilitate trauma recovery through lucid dreaming



NX-422 Neural Interfaces

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2 State-of-the-art review

Lucid dreaming (LD) is the experience of achieving conscious awareness of dreaming while still asleep. During a LD, the dreamer often gains a degree of control over the dream content and can voluntarily influence the dream narrative [37]. Lucid dreaming occurs spontaneously in about half of the population but rarely on a regular basis [60]. Recent research into LD has significantly advanced the understanding of consciousness, reality threshold and dream states, revealing its potential therapeutic benefits (source incoming). Notably, two-way communication between researchers and lucid dreamers has enabled real-time interaction during dreams [33]. To signal that they experience a LD, the dreamer can rapidly move its eyes resulting in a left-right-left-right pattern [5]. The is being explored as a promising treatment for recurrent nightmares, due for instance to a traumatic event [74, 13, 28]. Furthermore, the role of LD is being explored in enhancing creative problem solving [78], emotional regulation [67] or even motor enhancement [62].

Lucid dreaming induction techniques can be classified into four categories (1) cognitive techniques, (2) external stimuli, (3) substance intervention, and (3) cortical stimulation [81].

Cognitive techniques include the Mnemonic Induction of Lucid Dreams (MILD) technique, Wake Back to Bed (WBTB) method, and Reality Testing (RT). MILD involves repeatedly telling oneself before sleep that they will become aware of dreaming the next time they dream. It often includes waking up after about five hours of sleep, staying awake briefly, and then returning to sleep [36, 2]. In the WBTB method, the dreamer purposely wakes up early in the morning and then takes a nap, during which lucid dreaming becomes more likely [77, 2]. In RT, the dreamer regularly checks if they are dreaming throughout the day, creating a habit that hopefully carries over into their dreams [2]. The MILD technique is shown to effectively induce lucid dreaming, while WBTB and RT are proven less effective [81]. However, these techniques include the interrupting of sleep and consciously re-entering their dream, and oftens requires practice to successfully induce LD. Interrupting the Rapid-Eye Movement (REM) sleep can lead to parasomnia [9], or other negative feelings such as long term stress or anxiety [3], or insufficient or poor sleep [50].

External stimuli involves the application of sensory cues to make the dreamer aware that they are dreaming. Mainly auditive [17, 61, 69], visual [23, 37] or tactile cues [59] are used to warn the dreamer during sleep, allowing them to recognize the signal and subsequently taking control of their dream. Compared to the previous techniques, the application of sensory cues has the advantage of being able to be applied during REM sleep without prior practice, even if prior practice increase the probability of lucid dreaming [69]. However, it is important to not wake up the dreamer and ensure that the stimuli stay under a certain threshold, as well as not to alter their sleep quality [69]. This technique is used in both research-oriented and consumer-focused devices. Both DreamLight® [39] (now discontinued) and Hypnodyne's ZMax® [11] are specifically designed for sleep-related research. Hypnodyne's ZMax® is a portable sleep-monitoring headband with sophisticated features combining vibrotactile and auditory stimuli, EEG sensors and eye movement sensors, heart rate, temperature, light and sound monitoring, and customizable scripts, set apart by it's highly accurate REM detection algorithm [18]. However, it is costly, and while it demonstrates real accuracy, there is limited research or reviews available on its performance on the induction of LD, and is mainly used for sleep monitoring. On the customer side, devices are mainly in a headband format, and they use less accurate REM detection and attempt to induce LD in a similar manner as before. Currently on the market available devices include the Somni mask® [72], combining eye and head movement tracking, LED signals, and sound signals, as well as the iBand+® [29], using EEG brain sensing, pillow speakers, and audio-visual cues, and Remee [42], flashing light cues based on preset timers rather than real-

time sleep detection, making it less reliable at inducing lucid dreams compared to REM-specific cues. However, many devices using external stimuli have been discontinued for undocumented reasons, such as the NovaDreamer® [55], REM Dreamer [56], and Aurora Dream-Enhancing Headband by iWinks® [29]. The current commercial devices lack proof on how well they induce a LD and often require prior training on RT before using the device [46, 48]. According to a review study, using external stimuli alone have not been proven effective in the induction of LD, result are ambiguous and is often tested on subjects with prior LD experience [81]. Occasionally, the flashing lights or auditory stimuli produced by the device could even disrupt an ongoing LD.

Substance intervention, such as the administration of galantamine [75], can also be used to induce a LD. In this method, the dreamer wakes up after 4 to 5 hours of sleep and takes the acetylcholinesterase inhibitor galantamine (4 or 8 mg), which has been shown to significantly increase the frequency and quality of LDs while enhancing recall and sensory vividness [41]. However, it is important to administer the right dose at the right timing, and often is combined with cognitive techniques such as WBTB. Here again, accidentally interrupting REM sleep or changing the neurochemistry in the brain might result in undesirable sleep behavior [83].

Recently, the efforts regarding the induction of LD start to shift towards cortical stimulation. The LucidCatcher® [45], a commercial device, uses transcranial Alternating Current Stimulation (tACS) during REM to induce a lucid dream. While lower gamma frequency tACS is proven to be effective to some extend to induce lucid dreams [7, 90], the LucidCatcher® did not make it to market, potentially due to a safety concern for its customers [48]. Similarly, the Lucid Dreamer® applies tACS to induce a LD [15]. The testing of the prototype in 2017 showed promising results, with an induction of LD in 8 out of 18 participants, but has not been released ever since. The effectiveness of tACS in LD induction is ambiguatory, as the initial paper showing positive effects has not been replicated since [48, 90]. A new player in the field is Prophetic [63], which tries to induce LD by transcranial Focused Ultrasound (tFUS). It is hypothesized that the combination of a better understanding of the neural architecture underlying LD and more precise neuromodulation techniques such as tFUS potentially allow for a more specific and improved induction of LD [60]. However, there are no studies to our knowledge that provide concrete insights into the effect of tFUS on LD. Lastly, we noticed multiple sources indicating the positive effects of transcutaneous Direct Current Stimulation (tDCS) on LD [87, 76], but we found no devices currently using this technology for the induction of LD.

These methods indicate a broad spectrum of cognitive techniques, external stimuli, pharmacological techniques, and upcoming cortical stimulation devices with varying degrees of success. Bearing in mind the low success rate of commercializing a LD device, the major challenge here is the development and scientific validation of a reliable, safe technique that increases the occurrence of lucid dreaming [81]. Furthermore, current devices focus only on the direct induction of LD during REM sleep, while we believe that it is reasonable to look at LD from a complete sleep cycle point of view, combining both sleep enhancement stimulation as well as LD induction.

3 Our device

We propose a wearable multimodal closed-loop device that combines ear-EEG to monitor sleep stages, transcutaneous auricular Vagal Nerve Stimulation (taVNS) to facilitate REM sleep, and auditory stimuli and transcranial Direct Current Stimulation (tDCS) to induce LD. The novelty of our approach lies in the smart integration of all different components into a single device, ensuring seamless electrical integration, ease of use, comfort during sleep, and portability. In our device, tDCS will play the key role of inducing lucidity. However, it is key to apply these stimuli at the correct timing. Therefore, ear-EEG will facilitate the detection of REM sleep. To

improve the transition of the body from a wake to a sleep state and ensure proper sleep quality, taVNS is applied to the auricular branches of the vagus nerve. To bring this all together, we will briefly touch upon the role of smart REM detection algorithms and parameter tuning to ensure a personalized approach.

3.1 Biological background

To understand the biology behind LD, it is important to understand the normal sleep cycle, and the brain activity during dreaming. Usually, the sleep-wake cycle in humans can be divided into three alternating objectively identifiable states of consciousness of the brain [26]: waking state, non-rapid eye movement (NREM) sleep and REM sleep, as depicted in Figure [..]. The three states can be detected through measuring electroencephalography (EEG) signals. The wake EEG is dominated by low-voltage (5–50 µV) fast-activity in the beta (16-30Hz) and gamma (>30Hz) range [8], relative to voltages and frequencies seen in the NREM sleep stage. When transitioning from wake to NREM sleep and progressing toward deeper sleep stages 1, 2, 3, and 4, the amplitude of the dominating brain waves increases (100–400 µV), while the frequency decreases (0.5–3 Hz), which are called delta waves [71]. When entering REM, the EEG signal exhibits similar oscillations to the wake stage, but can be distinguished from it due to the presence of phasic events, such as rapid eye movements [8]. In addition, the amplitude is higher, and the frequency lowers, resulting in theta waves (4-7Hz) [71].

Although brain activity during REM resembles that to the wake stage, the cognitive processes during REM sleep are heavily constrained. This is primarily due to the deactivation of the dorsolateral prefrontal cortex (dlPFC) [27, 47]. The dlPFC lies within the prefrontal cortex and is central to working memory and abstract reasoning [91], as well as self-focused metacognitive evaluation [84]. Interestingly, neuroimaging data show increased activity in the dlPFC during LD [16, 89]. LD is an intermediate position between REM sleep and waking, and consists of both waking and dreaming features. Subjects regain many aspects of waking while continuing to dream [38], they become aware of their dreaming state, have full access to memory, and are able to volitionally control dreamed actions. Moreover, the transitioning from non-lucid to lucid REM sleep is associated with an increase in the high-frequency gamma (40Hz) band of their frontal and frontolateral EEG [89], denoting synchronization of cortical neuronal activity essential for the waking state. The power in the delta and theta band remain similar to the REM state.Therefore, our device applies stimulation to the dlPFC during REM sleep in the form of tDCS. By applying a direct current through the dlPFC, we can increase its neuronal activity [53, 54]. This means that the cognitive functions essential for LD are restored, resulting in effective LD induction [10, 76, 70].

It is hypothesized that deep and restful sleep results in increased REM time in the early morning, resulting in more vivid dreams [83]. In addition, the quality of sleep is an incredibly important, yet often overlooked aspect of LD. Poor quality of sleep, in combination with a physically, mentally, or emotionally exhausted state of mind is undesirable for LD and may result in adverse effects for our health [50]. Bearing in mind the desirable mental and physical states of when to apply LD induction, we stimulate the auricular branches of the vagus nerve using taVNS, which originates from the ganglion of the vagus nerve from within the jugular foramen [92, 82]. By triggering brain regions involved in ascending sleep-related projection, including the locus coeruleus, periaqueductal gray, hypothalamus, and thalamus, taVNS induces the secretion of melatonin and inhibitory γ-aminobutyric acid (GABA) [4] resulting in a smooth transition to the sleep stage and directly improving sleep quality and behavior [12, 66, 51]. Furthermore, the excitability of the hypothalamic-pituitary-adrenal axis decreases, alleviating stress symptoms [25]. By applying taVNS, a significant increase in delta oscillations is found, proving that taVNS effectively promotes the transition to (deep) sleep stages [22, 6].

3.2 Fabrication & encapsulation

Our device is designed to be wearable and multi-modal, making it ideal for at-home use. To achieve this, we have developed a prototype that combines taVNS and tDCS stimulation with ear EEG functionality. The device will be crafted from silicone and custom-shaped to fit the user's ear, ensuring both precise measurements and maximum comfort. The design process involves using 3D software to model the device based on the user's head measurements, creating a mold from this model, and then casting the final silicone resin using the mold. All 2D design can be found in the Appendix (Section 6).
The design is as follows:

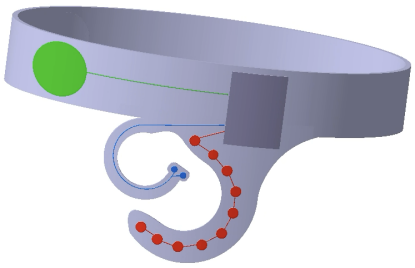


Figure 1: 3D design of our prototype

3.2.1 tDCS

Two C/Ag/AgCl electrodes [43], functioning as an anode and a cathode, are used to deliver a stimulation current through the skin to modulate brain activity, here in green in Figure 1. These electrodes create a closed loop that ensures a consistent current flow to the stimulation sites in the brain. While the applied electric current does not generate action potentials, it modify ongoing brain activity, producing effects on brain function that extend beyond the stimulation session.

The electrode size is 5 cm in diameter. This is determined by the current requirements for home use and is tailored to safely and effectively deliver the desired stimulation. This size is specifically chosen to distribute the current over a larger surface area, reducing the current density and minimizing the risk of discomfort or tissue damage at the electrode-skin interface.

These C/Ag/AgCl electrodes can be fabricated using a cold rolling [43] process that involves coating a carbon layer onto Ag/AgCl electrodes. This method takes into account the final electrode dimensions, eliminating the need for microfabrication techniques. Carbon electrodes are ideal for tDCS because they are lightweight, adaptable, and highly resistant to wear and corrosion. Additionally, their inert nature minimizes the risk of skin irritation, ensuring safe and comfortable use during stimulation sessions.

3.2.2 taVNS

For transcutaneous auricular vagus nerve stimulation (taVNS), two Iridium-titatnium coated electrodes are used to deliver the stimulation of the auricular vagus nerve through a non invasive placement of the electrodes on the cymba conchae, see in blue in Figure 1. As seen in the biological background, taVNS is said to increase the amount of REM sleep, enhance the self-awareness in dreams and improve dream recall [12, 66, 51]. Additionally, it also facilitates the body's transition to a sleep state.

Our Transcantaneous choice, which avoids breaking of the skin for the electrodes placement, is justify by an easy use for a at home device. It can be self-administered and has a minimal chance of infection or significant side effects. This makes it a more practical choice compared to the percutaneous method, which, while offering more precise stimulation, requires medical placement and is less ideal for commercialization.

The electrodes required for taVNS are 2 mm in diameter. Iridium-titanium electrodes are fabricated by electrodepositing iridium onto a titanium substrate [94] which is often used because of its excellent mechanical properties and resistance to corrosion. During fabrication, a titanium substrate is immersed in an electrolyte solution containing iridium ions. A voltage is applied

between the titanium substrate (serving as the cathode) and another electrode (typically the anode), causing the iridium ions to reduce and deposit onto the titanium surface, forming a coating layer. This microfabrication technique is selected because it achieves the required electrode thickness in the millimeter range [34].

3.2.3 Ear EEG

The ear EEg is a C-shaped electrode array as per the cEEgrid [14] and is specifically designed to be placed around the user's ears, as you can see in red in the figure 1. It is 3D modeled for the user ear and then made of silicon. This design choice aligns with our goal of creating a personalized, at-home device. Traditional EEG systems can be uncomfortable for users, especially when used during sleep. Moreover they need medical assistance to be put in place. As an in-ear EEG is more compact, its placement would interfere with taVNS stimulation and refrain the user from hearing around which can be unsafe and impractical. It is also less precise, as it is not in direct contact with the brain but is placed on the ear [31] . An ear EEG positioned around the ear is an ideal alternative, as it avoids hair interference, reduces contact resistance and noise, and remains highly discreet and comfortable. Moreover, this placement allows the device to maintain direct contact with the brain's electrical activity while leveraging the ear as a natural anchoring point for secure attachment.

The array is constituted of 10 electrodes: 8 measuring ones as well as a reference one and a ground one. The size of each electrode is 5 mm diameter and they are made in Ag/AgCl because they are well suited for low frequency signal recording [35]. We need to apply a little bit of gel to assure a good contact, so our ear EEG is not entirely dry. Here is the microfabrication flow of these electrodes: [14].

1. Start with a polyamide substrate
2. Use a shadow mask to define the electrode area
3. Deposit a 100 nm titanium (Ti) layer as an adhesion layer and 550 nm silver layer on top of the Ti layer using E-beam evaporation
4. Removal of the mask
5. Immerse the silver layer in $FeCl_3$. It converts the outer layer of Ag to AgCl based on the reaction: $Ag(s) + FeCl_3(aq) \rightarrow AgCl(s) + FeCl_2(aq)$
6. Seal the pool to isolate it from the external measurement solution.

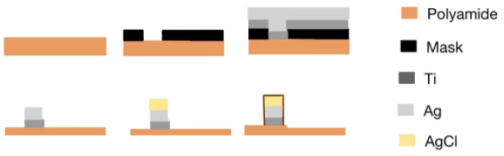


Figure 2: Microfabrication flow for the Ag/AgCl electrodes for the ear EEG

3.3 Electrical system architecture

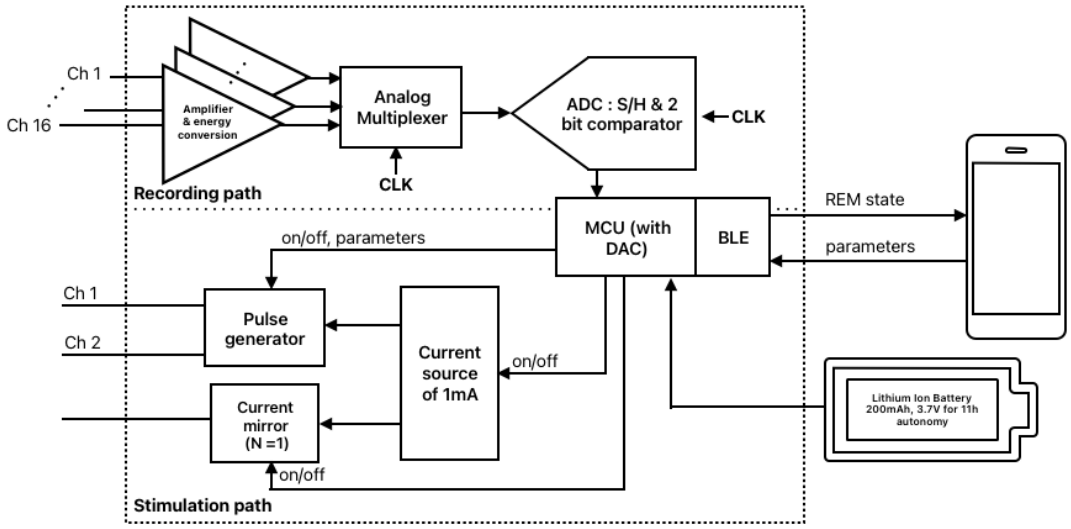


Figure 3: Block diagram of the system architecture

3.3.1 Amplification

The recording path begins with a 16-channel EEG signal (each ear has a ground and a reference electrode) which typically ranges up to 50 μV [49]. This signal is first amplified and filtered. To isolate the frequency band associated with REM sleep (which spans 15–35 Hz [88]), a capacitive AC amplifier is used. The transfer function of this amplifier is given by:

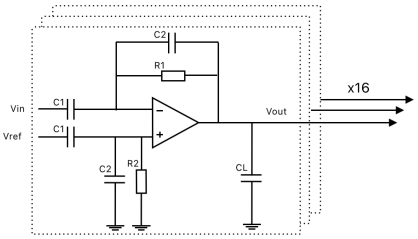


Figure 4: AC amplifier

$$\frac{v_{\text{out}}}{v_{\text{in}+} - v_{\text{in}-}} = A_M \cdot \frac{1 - s/(2\pi f_z)}{\left(\frac{2\pi f_L}{s} + 1\right)\left(\frac{s}{2\pi f_H} + 1\right)},$$

Where $f_L = \frac{1}{2\pi R_2 C_2} = 15\,\text{Hz}$ is the lower cutoff frequency, $f_H = \frac{G_m}{2\pi C_L A_M} = 35\,\text{Hz}$ is the higher cutoff frequency, and $f_z = f_H \cdot \frac{C_1 C_L}{C_2^2}$ is the zero frequency. The gain of the amplifier (A_M) is determined by the ratio of the capacitors C_1 and C_2 : $A_M = \frac{C_1}{C_2}$. The amplifier thus acts as a band-pass filter, amplifying only the REM-relevant frequencies in the bandwidth equal to : $BW = 35\,\text{Hz} - 15\,\text{Hz} = 20\,\text{Hz}$, while attenuating the others.

3.3.2 Energy conversion

We look at the energy of the signal because the amplitudes of brain signals are too unstable to carry reliable information about REM sleep stage. Therefore, one use a full wave rectifier that converts the AC signal into an all-positive signal to avoid cancellation during squaring. Then, we square the rectified signal giving us the power as it is proportional to the square of amplitude. Finally, the use of a low pass RC filter allows to average the energy by smoothing the squared signal over a short time window.

3.3.3 Multichannel analog to digital conversion

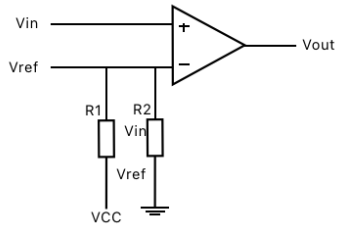


Figure 5: Non inverting comparator

The amplified EEG power signals are fed into a multiplexer. Indeed, speed is not crucial in our application, therefore using a multiplexer allows us to gain area and power by having only one analog to digital converter (ADC) in the whole device.
Our ADC is a simplified successive-approximation-register (SAR). The difference is that we do not need a complex adaptive threshold because we only want to know if there is a high enough amount of energy, we are in REM sleep and $V_{\text{out}} = 1$, otherwise we are not in REM sleep stage and $V_{\text{out}} = 0$.

The voltage threshold is simply set by a voltage divider from V_{cc}. Therefore, after being sampled and holded at a frequency of $f_s = 2 \times 35Hz = 70Hz$, the output is as followed :

$$V_{\text{out}} = \begin{cases} 1, & \text{if } V_{\text{in}} > V_{\text{threshold}}, \\ 0, & \text{if } V_{\text{in}} \leq V_{\text{threshold}} \end{cases}$$

Since we only have a 1 bit resolution, we don't need any compression of the data.

3.3.4 Transmission and control

The control code will be on the micro-controller (MCU) chip. The details on data transmission and the control algorithm are specified in this part here. The MCU have a bluetooth low energy (BLE) unit as well as a digital to analog converter (DAC). The BLE is used to transmit the state (REM or not REM) to a phone application (more details here) and to receive some parameters from it. The parameters are meant to personalize the taVNS current. The MCU controls the stimulation path with switches made of MOSFET mainly because of their good power efficiency their scalability and their fast switching speeds.

3.3.5 Stimulation path

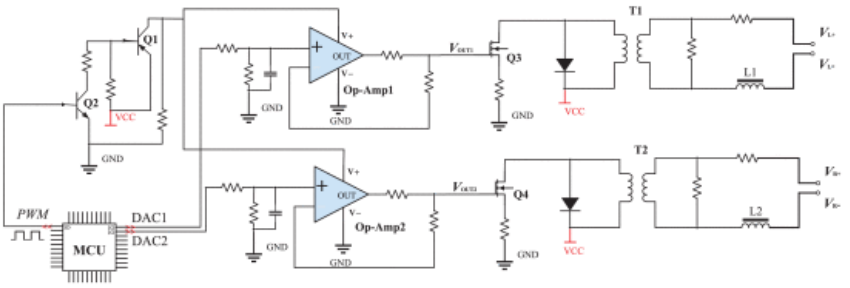


Figure 6: Pulse generator circuit [95]

The main component of the stimulation path is the pulse generator [95]. It consists of two parallel circuits aimed at producing balanced current pulses in both ears to stimulate the vagus nerve. First, there is a current source of 1mA implemented with the bipolar transistor Q_1, whereas Q_2 is used to amplify the circuit in the saturation region. The signals from the MCU are pulse width modulated (PWM) signals and two DAC signals (DAC1 and DAC2), which respectively control the frequency of the pulse, as well as its amplitude and pulse width. Afterward, the Q_3 and Q_4 NMOS transistors provide additional electrical insulation. The flyback diodes prevent back electro-motive force (EMF). Finally, the transformers T_1 and T_2 are used to step up the voltage to overcome the skin resistance. This setup creates the balanced current

pulses for both taVNS electrodes, which typically operate with a frequency $f \approx 20$ kHz, current $I \approx 2$ mA, and pulse width $PW \approx 20$ µs. For tDCS stimulation, only a current of 1mA is required. Therefore, a current mirror can be used from the current source in the pulse generator with a size ratio of $N = 1$.

3.3.6 Preliminary tuning

The tuning process depends on numerous factors. While this example represents a single trial, determining the optimal tuning requires extensive experimental testing and solving optimization problems. Within the scope of this project, we tried to achieve this using the tools available to us.

Gain Design : By choosing a gain of 1000, the amplified EEG signal ranges from 0 mV to 50 mV. Specifically, the gain A_M is given by $A_M = \frac{V_{\text{out}}}{V_{\text{EEG}}} = \frac{50\,\text{mV}}{50\,\mu\text{V}} = 1000$, where V_{out} is the amplified voltage and V_{EEG} is the maximum amplitude of the EEG signal. This gain can be achieved in a single stage with $C_1 = 1000\,\text{pF}$ and $C_2 = 1\,\text{pF}$, or alternatively, it can be split across two amplifiers to improve stability and bandwidth performance.

Energy circuit design : After rectification, the signal remains in this range. Squaring converts the signal to a range of $0\,\text{mV}^2$ to $(50)^2 = 2500\,\text{mV}^2$. A low-pass filter then calculates the average energy, which for a sinusoidal signal is proportional to the RMS value. The RMS of a 50 mV peak signal is $A_{\text{RMS}} = 0.707 \cdot 50 = 35.35\,\text{mV}$, giving an average squared value of $A_{\text{RMS}}^2 = 1250\,\text{mV}^2$. Assuming that the circuit scales $1000\,\text{mV}^2$ to 1 V, the low-pass filter output is approximately up to 1.25 V. Thus, the final voltage range is 0 V to 1.25 V.

Threshold voltage : The reference voltage $V_{\text{threshold}}$ represents the threshold for detecting REM activity. It is set by a voltage divider from V_{cc} which is equal to 3,3V. Let us set this voltage at half of the signal range $\frac{1.5V}{2} = 0.75V$. Therefore we can use $R_1 = 400\ \Omega$ and $R_2 = 118\ \Omega$, allowing to have $V_{\text{threshold}} \approx 0.75$V as follow:

$$V_{\text{threshold}} = V_{\text{cc}} \times \frac{R_2}{R_1 + R_2} = 3.3 \times \frac{118}{118 + 400} \approx 0.75V$$

Noise considerations : It is important to consider the noise floor of the sensory circuit. Based on ear EEG studies, a noise spectral density of $122\,\text{nV}/\sqrt{\text{Hz}}$ [58] is adequate when recording. The total RMS noise voltage V_{RMS} over the bandwidth is the noise spectral density times the square root of the bandwidth:

$$V_{\text{RMS}} = \text{Noise Density} \times \sqrt{\text{BW}} = 122\,\text{nV}/\sqrt{\text{Hz}} \times \sqrt{20\,\text{Hz}} \approx 545\,\text{nV RMS}$$

Power considerations : The following table presents the different entities of the neural interface, an example component for each block, and the estimated power consumption in watts assuming a supply voltage of $V_{\text{CC}} = 3.3$V and based on their average current.

Component	**Example Part**	**Power (mW)**
Amplifiers (16x)	AD8606 (10 µA each)	0.528
Energy Conversion (16x)	LTC1966 (20 µA each)	1.056
Analog Multiplexer	CD4051 (50 µA)	0.165
Comparator	LM393 (60 µA)	0.198
MCU with BLE	STM32L432KC	33.7
Stimulation circuit	–	≈ 5
Total	–	$\approx$ **40mW**

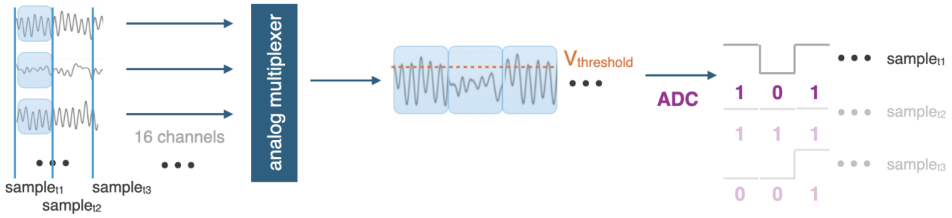
Table 1: Power consumption breakdown of the neural interface.

The total estimated power consumption for the neural interface is approximately 40 mW with a supply voltage of $V_{CC} = 3.3\,\text{V}$. Since we made several approximations we add a margin of 10mV, therefore our total power is P = 50mW. For an autonomy of 11h, which will cover a sleep cycle, we will use a lithium ion battery of 150mAh capacity and with a nominal tension of 3,7V, as shown by the following calculations.

$$E_{\text{total}} = P \times t = 0.05\,\text{W} \times 11\,\text{h} = 0.55\,\text{Wh}$$

$$\text{Capacity}\,(mAh) = \frac{E_{\text{total}}\,(\text{Wh})}{\text{Tension}\,(V)} \times 1000 = \frac{0.55\,\text{Wh}}{3.7\,\text{V}} \times 1000 \approx 149\,\text{mAh}$$

3.4 Data Processing Pipeline



The data processing pipeline is denoted here in greater depth. It begins with 16 normalised voltage transients from the two ear EEG devices. Our multichannel architecture necessitates an analog multiplexer, which extracts 16 synchronously occurring samples, and concatenates them together into one signal for the ADC.

The objective is to aggregate the signals across all channels, and eventually come to classify the sleep stage as being in or not in REM. The signals have been filtered such that the beta signals, which characterize REM sleep, remain. Therefore, we threshold this filtered and concatenated single signal to some voltage threshold. That is to say that if the strength of beta frequency signals across all channel reading is strong enough, the device classifies the sleeper as being in REM.

You can see here that the ADC output is a 16 bit string, where more 1s indicate more REM power. This is an input into the MCU's control code.

Through a smartphone app, users will be able to configure the parameters for their stimulation, including the duration and intensity. This flexibility will be capped as per safety restrictions.

In line with the potential therapeutic benefits of lucid dreaming, the mobile application will also offer dream journaling functionality, a practice which improves one's ability to control dreams [40]. This interface also offers private and secure access for the subject's psychologist, therefore enabling the use of lucid dreaming as a means for therapy and trauma healing [57]. Users will also have overnight and historical recounts of the times that tDCS was triggered.

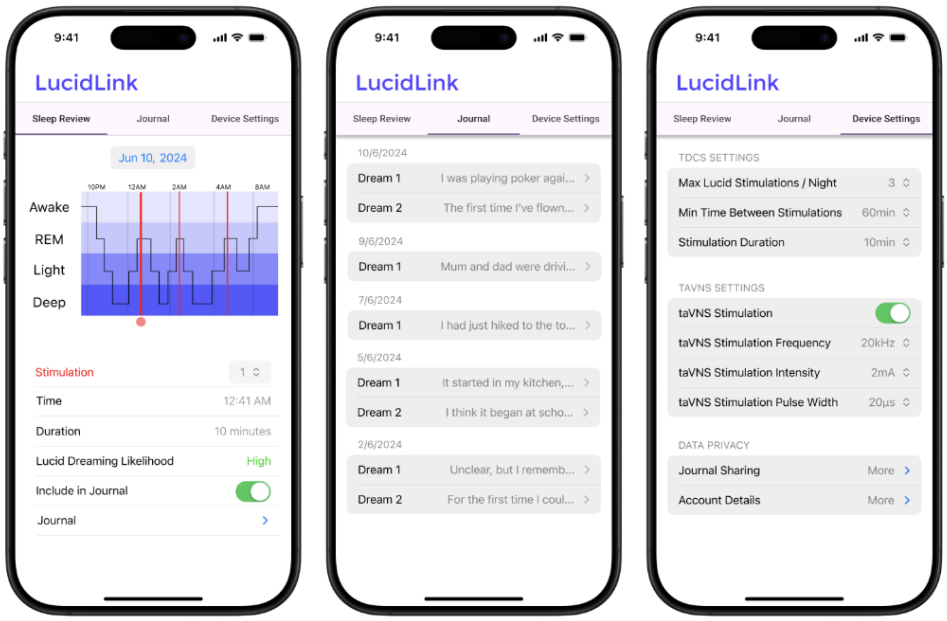


Figure 7: iOS Application UI

```
internal input: sample # string of 16 bits
external input: dur_rest, dur_taVNS, dur_tDCS # durations of stim & rest
f_taVNS, intensity_taVNS, PW_taVNS
output: (trigger tDCS, taVNS freq, taVNS intensity, taVNS PW),
trigger current source

counter   <- counter_t-1 + 1
is_REM    <- is_REM_t-1      # retain from previous run
strengths <- strengths_t-1[-12000:].append(mean(sample))
curr_strength <- mean(strengths)  # ∈ [0,1] # memory of 2min

if is_REM and (curr_strength < thresh_low or counter >= dur_tDCS):
      # stop tDCS due to insufficient REM or elapsed time
      is_REM <- 0
      counter <- 0
      to_return = 0, f_taVNS, intensity_taVNS, PW_taVNS
else if is_REM:
      to_return = 1,0,0,0
else:
      if counter > dur_rest and curr_strength > thresh_high:
            # recommence tDCS
            is_REM <- 1
            counter <- 0
            to_return = 1,0,0,0
      else if counter < dur_taVNS:
            to_return = 0, f_taVNS, intensity_taVNS, PW_taVNS
      to_return = 0,0,0,0

return to_return, if(to_return) # if(to_return) => current src
```

Figure 8: MCU Source Code

The control code required for the MCU chip is easily hardcoded. Its reliance on transmitted parameters from the smartphone device means removes any need for software updates on the algorithm.

As detailed on the left, The algorithm receives a 16-bit string and several configured parameters. The output is a series of integers detailing whether tDCS or taVNS is to be on (and with what waveform), and therefore whether the current source is required. This is decided through some `counter` variable which keeps track of time, ensuring that stimulation is temporally restricted, and the `rest` periods desired by the user elapse before recurrent stimulation is induced.

The strength of the REM signal is derived from the distribution of 1s in the recently inputted bit strings. Switching the truth of `is_REM` from `0` to `1` is done only if the weighted average of the previous 2 minutes of activity is above $\texttt{thresh}_{\texttt{high}}$. The converse is true if the weighted average is below $\texttt{thresh}_{\texttt{low}}$.

4 Feasibility assessment

The manufacturing of our device with the materials outlined is expected to be largely feasible. That said, there are some risks surrounding the function of the device given our decision to create a design prioritizing subject comfort.

For example, literature backing use of the more comfortable carbon electrodes for the tDCS component exists [43] but is limited. Carbon electrodes have lower conductivity than their electric counterparts and thus may result in larger impedance. Alternatives to this include Mul-

tilayer Hydrogel Composite (MHC) dry electrodes [32] or Flow's tDCS device's metal electrodes with replaceable saline solution pads [20].

Similarly, dry ear EEGs show promising accuracy [44] yet suffer from the impedance caused by the absence of an electrolyte.

The REM detection algorithm may present some flaws during the testing period. EEG signals in the REM stage, which are explored in Section **??**, are characterized by an increase in the power of the beta (16-30Hz) and theta (4-7Hz) oscillations. Our device focuses on detecting large voltage deviations in the beta waveband, but REM detection would be more precise by converting EEG signals into a power spectrum density and identifying larger powers in the mentioned frequencies. This approach would require a delta-sigma ADC [24] and possibly an off-chip data processing algorithm, largely increasing the device's area and power demands. Dynamic machine learning models have successfully decoded EEG signals in the past [68, 1, 65] but not yet specifically for the purpose of lucid dreaming. Future alterations made to the device could record neural activity near the frontal cortex, which offers more reliable signals [89].

Adding cloud-reliant algorithms to the sleep detection process depends on the region-specific data privacy laws. In Switzerland, biometric data is to be treated as sensitive data [19] and other countries hosting such data must agree with protocols [30].

The specific regulatory standards our device would need to pass is made more complicated by its multi-component nature. Attaining approval for the stimulating components of the device (tDCS and taVNS) would prove the biggest hurdle Considering first America's FDA, our device would likely be considered a Class II Medical Device, in line with other non-invasive electrical stimulation apparatuses.

Approval is typically achieved through the submission of a 510(k) form [85] which requires a reference to a commercially available device of equivalent technology and similar purpose. Given the novelty of our device, attaining approval would prove difficult and would necessitate the submission of a De Novo Request [86], which in turn requires data from "bench performance testing". Indeed, clinical trials were needed for the aforementioned Flow device[52].

Elsewhere, regulatory standards for tDCS devices are relatively unclear between regions [21]. In September 2023 the first tDCS device was approved for the European market, having been deemed conformant with EU Medical Device Regulation (MDR)[73]. Precedent for tVNS devices is less clear - the technology is more naissant [93] - but tVNS devices have also previously been approved under MDR [79] for the treatment of neurological diseases.

Potential consumers of our device may be reluctant to stimulate their cranium during subconscious sleep, even at low amplitude and a temporally limited timeframe. Furthermore, previously unsuccessful lucid dream devices are reviewed poorly for their impedance of sleep[**?**].

5 Conclusion

Our wearable multimodal device represents a novel approach to inducing lucid dreams by integrating ear-EEG, taVNS, auditory stimuli, and tDCS into a single, configurable, and comfortable system. Although much thought has been spent creating a lightweight and comfortable design, the device may prove disruptive to some in practice. Future iterations could improve by adopting wireless designs [64, 91] or single-ear EEG setups [80]. One could also add an inertial measurement unit to correlate the movement of tze body with the sleep quality, offering a better sleep analysis.

talk about our innovation being the multi-modal aspect, the configurability, and the comfort. Talk about the need to validate the algorithm / EE logic in the testing of the device (ie will we need to use the brainwaves PSDs?)

6 Appendix

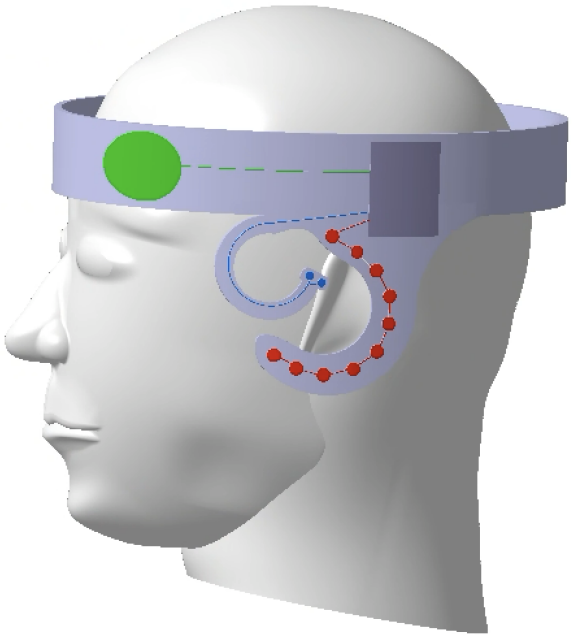
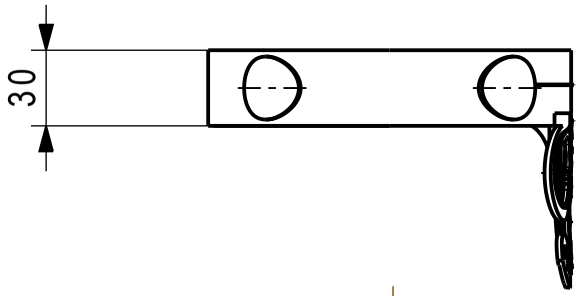
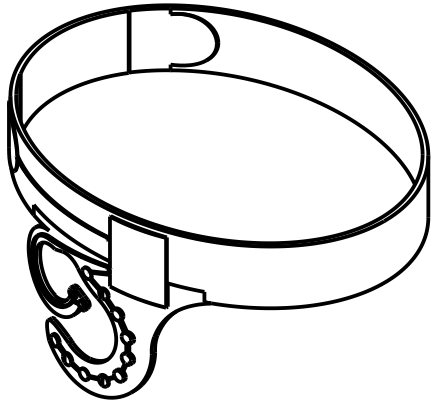
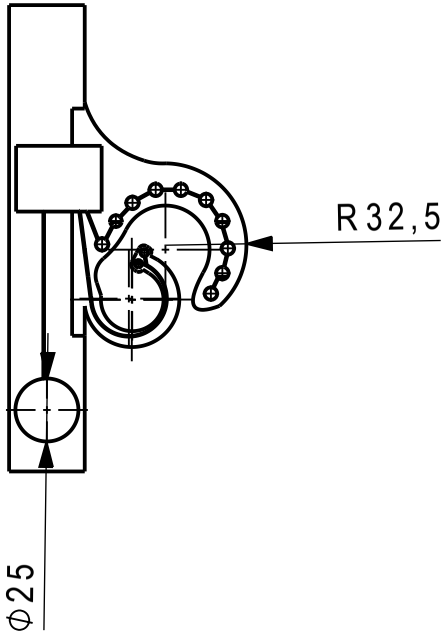
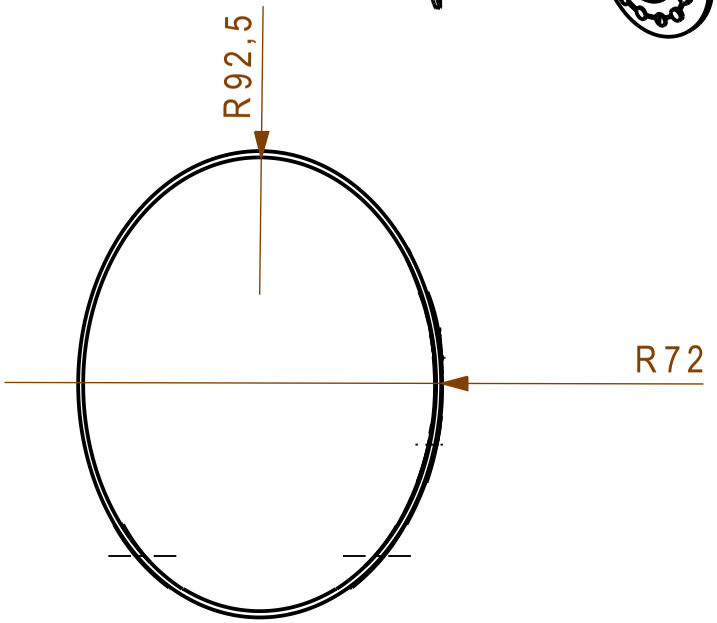


Figure 9: 3D model worn of our prototype









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				Contrôlé		
				Conf aux norm		
				Bon pour exéc.		

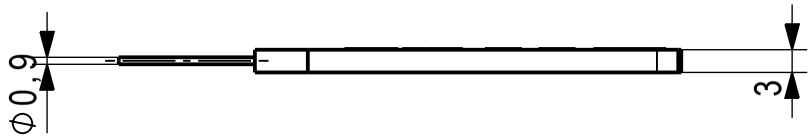
Echelle 1:3

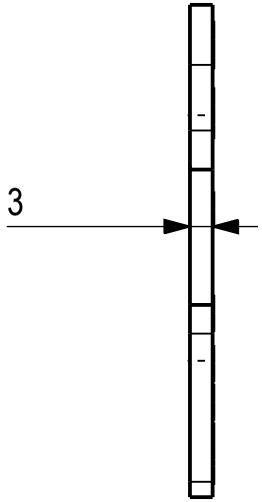
Sans nomenclature séparée ☐			N° de commande		
Nomenclature sép de même N° ☐	Matiere	.	Origine		Format A4 / Nb feuilles 1 / Feuille N° 1
Nomenclature sép de N° diff ☐	Masse	kg	Remplace		

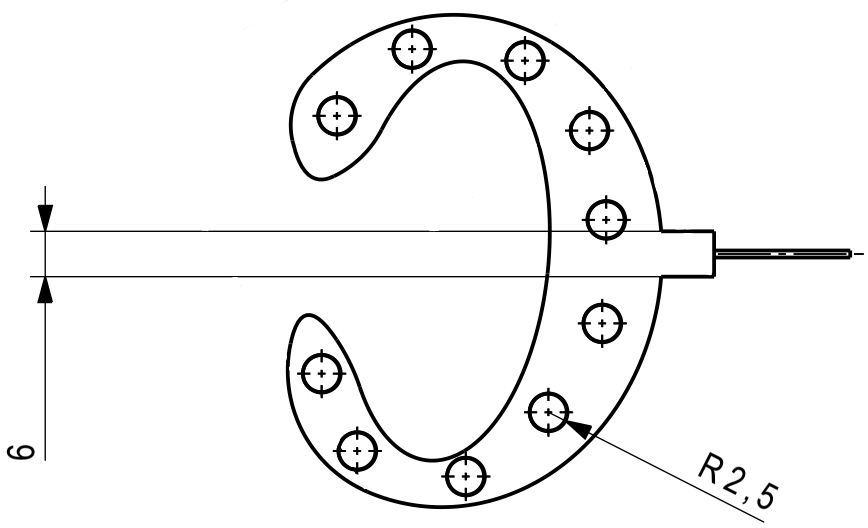
EPFL

Dénomination: LucidLink

N° de dessin







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1	10		electrodes ear	

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Contrôlé			
Conf aux norm			
Bon pour exéc.			

Mod.

Mod.

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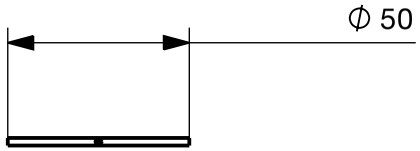
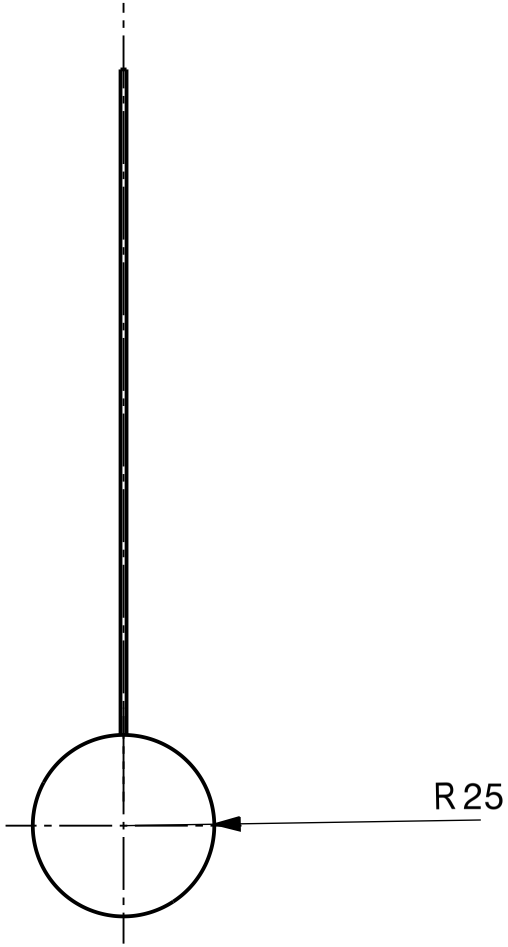
EPFL

Dénomination

Ear EEG

N° de dessin





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Nomenclature sép de N° diff ☐		Masse		Remplace		

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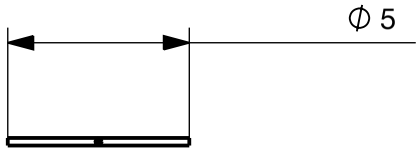
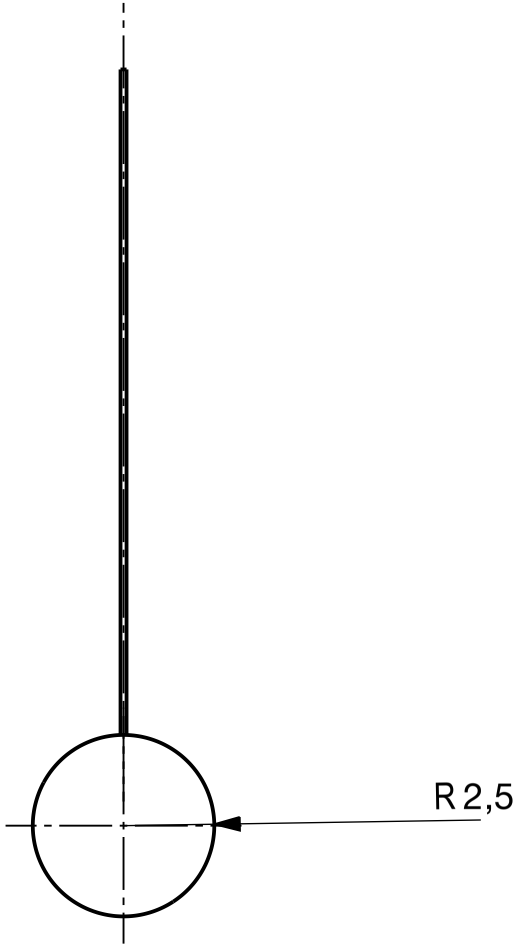
EPFL

Dénomination: TDCs

N° de dessin





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Echelle 2:1

Format A4 | Nb feuilles 1 | Feuille N° 1

EPFL

Dénomination

EAR EEG ELECTRODES

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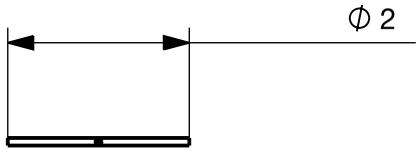

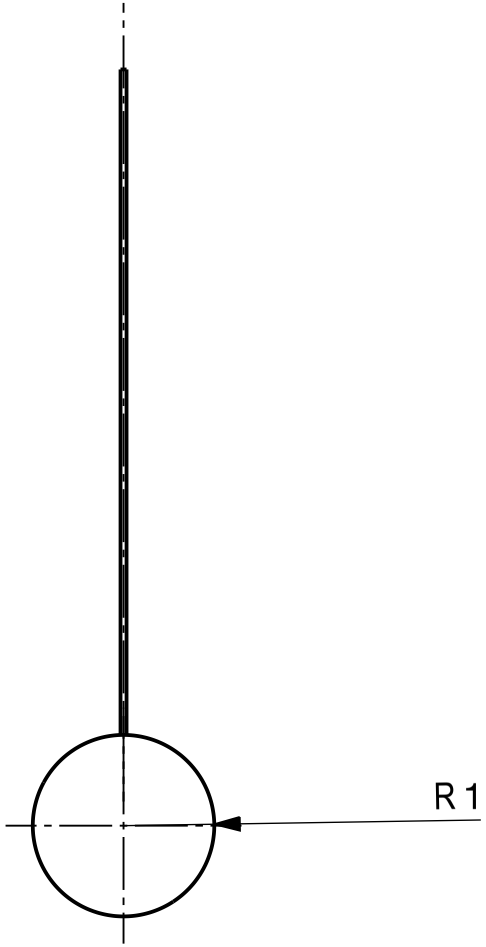

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EPFL

Dénomination

TAVNS

N° de dessin

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