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## Abstract

The dependency of a novel sensory feedback for stump-socket based prosthesis on the static force is presented using a bone conduction transducer as feedback source. The stimulation was induced onto the bony landmarks of the elbow, specifically the Ulna and presented in an interval halving method. The perception threshold in the range of tactile and auditory perception at three different force levels has been tested. The inter subject variability is bigger than the intra subject variation. The small static force variation suggests a similar approach as in bone conduction hearing aids and therefore a static force bigger than 6N should be applied to perceive a constant stimulation. A mechanical design to include such a novel feedback into a stump-socket needs to account for this requirement. The inter subject variability needs to be addressed by incorporate some kind of person to person calibration of the gain.

## Disciplines

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# Static Force Dependency of Bone Conduction Transducer as Sensory Feedback for Stump-Socket based Prosthesis

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## Abstract

The dependency of a novel sensory feedback for stump-socket based prosthesis on the static force is presented using a bone conduction transducer as feedback source. The stimulation was induced onto the bony landmarks of the elbow, specifically the Ulna and presented in an interval halving method. The perception threshold in the range of tactile and auditory perception at three different force levels has been tested. The inter subject variability is bigger than the intra subject variation. The small static force variation suggests a similar approach as in bone conduction hearing aids and therefore a static force bigger than 6N should be applied to perceive a constant stimulation. A mechanical design to include such a novel feedback into a stump-socket needs to account for this requirement. The inter subject variability needs to be addressed by incorporate some kind of person to person calibration of the gain.

## 1 Introduction

Current available active powered, myoelectrically controlled transradial prosthesis allow the user to accomplish many activities of daily living providing a functional restoration of the main grasps [Farina and Amsüss, 2016]. However, no commercially available interface for sensory feedback is yet available though highly desired by prosthesis users [Hartmann *et al.*, 2015][Cordella *et al.*, 2016].

Key aspects of successfully controlling an active powered upper limb prosthesis is the involvement of both the efferent (motor control) and afferent (sensory feedback) pathway. Currently available active powered upper limb prosthesis (OttoBock Michelangelo Hand, Bebionic 3, touch bionic i-Limb quantum, Vincent

Systems, VINCENTevolution 3,...) willingly only implement the motor control path [Belter and Dollar, 2011][Belter *et al.*, 2013]. The only commercially available prosthesis providing sensory feedback as a vibration symbolizing the grasping force was the VINCENTevolution 2 (available until 2017)[Svensson *et al.*, 2017]. Providing intentional sensory feedback is therefore a largely unresolved issue [Stephens-Fripp *et al.*, 2018] [Antfolk *et al.*, 2013] [Svensson *et al.*, 2017].

The sensory feedback interface can be categorized into invasive and non-invasive approaches. Non-invasive approaches covering electrotactile, vibrotactile, mechanotactile, audio, temperature and hybrid feedback mechanism. Where as invasive methods are targeted sensory reinnervation (TSR), peripheral nervous system stimulation (PNS), central nervous system stimulation (CNS) [Svensson *et al.*, 2017][Stephens-Fripp *et al.*, 2018]. In [Svensson *et al.*, 2017] the limitations of the single methods show the main limitation of the current available interfaces ranging from bulky and power consuming for mechano- and vibro-tactile feedback, unpleasant feeling for electrotactile feedback, recovery time for TSR and short life time for PNS and CNS.

Differing results have been obtained concerning the improvement of the actual control task due to the inclusion of feedback mechanism, though it is believed that feedback mechanism is essential for feeling body ownership of prosthesis and therefore reducing the currently high prosthesis rejection rate of reported up to 19 to 39% for upper extremity prosthesis [Stephens-Fripp *et al.*, 2018] [Svensson *et al.*, 2017] [Schofield *et al.*, 2014] [Antfolk *et al.*, 2013]. [Farina and Amsüss, 2016] state that non-invasive sensory stimulation will likely become common part of clinical prosthesis in the medium term while [Schofield *et al.*, 2014] points out, that the achieved feedback signal must input the correct stimulus in a non distractive manner.

In [Clemente *et al.*, 2017] it was demonstrated, not only tactile but also auditory pathways are involved in the effect of osseoperception in bone-anchored prosthesis, transmitting mechanical vibration via bone conduction, and that such a sensory feedback device could enhance the sense of ownership of the prosthesis and therefore improve quality of life of the amputees. Applying mechanic vibrations in the range of tactile as well as auditory perception, to the bony landmarks of the elbow, in a non-invasive manner therefore represents a novel sensory feedback interface for amputees. Before the limitations of such a interface can be investigated, basic design requirements, known in the field of bone conduction hearing aid devices for the skull have to be investigated.

This paper presents an investigation on the static force dependency of such a novel sensory feedback interface. In applying vibrotactile feedback to a person's limb with the goal of bone conduction, it is necessary for the transducer not only to be placed in contact with the (bone/skin) but also to be held firmly against it for achieving a constant acoustic impedance in order for sensation to be appropriately perceived. In conventional sockets a constant force applying the transducers to the bony landmarks of the amputee, also referred to as static force, cannot be guaranteed over a large range due influences like residual limb volume fluctuations at different times. Causes in the post-operative time (12-18 month) are edema (from surgery and/or injury), post-operative muscle atrophy discrete, post-operative fluid collections and residual limb muscle activity. Daily fluctuations are due to comorbidities, prosthesis fit, activity level, ambient conditions, body composition, dietary habits, and for women, menstrual cycle [Sanders *et al.*, 2012]. Ranges of in session fluid volume changes have been shown by [Sanders and Fatone, 2011] to be in the range of -8.5%/h to +5.9%/h whilst between sessions ranged from -2.6%/h to +1.2%/h within 3 to 5h.

This investigation therefore will provide first design requirements for the inclusion of bone conduction transducers into a stump-socket to deliver effective sensory feedback. Therefore, it is yet to be investigated if and to what extend such a novel sensory feedback interface is dependent on the static force by changing the attenuation of the bone conduction and therefore resulting in a variation of the felt sensation amplitude and threshold. Based on these results future work can then proof the possibilities of such an interface investigating parameters like temporal and spatial resolution as well as improvements of control tasks and the increase of body ownership.

After an introduction and explanation of the background of osseoperception and bone conduction in section 2. In section 3, the experimental setup as well as the calibration of the force sensitive resistors and the test procedure is explained. Section 4 shows the obtained test results and section 5 gives a final conclusion of the obtained results and a future outlook.

## 2 Background

To connect a prosthesis to the amputees stump, the conventional method is using a stump-socket which is partly covering the limb. Stump-socket technology is only applicable if a long enough stump is available, for short stumps of trans-humeral amputees it blocks shoulder movement and for trans-radial amputees it limits the elbow movement [Farina and Amsüss, 2016]. Recent advances have introduced the method of osseointegration as a new feasible mechanical prosthesis interface. By implanting a fixture into the bone on the one side and penetrating the skin on the other. The prosthesis is then connected to an Abutment which is screwed into the implanted Fixture [Li and Brånemark, 2017]. In osseointegration a sensation called osseoperception is present. Osseoperception, is a sensation due to mechanical stimulation [Clemente *et al.*, 2017]. Recent research has shown that touch and hearing are involved in this phenomenon and therefore it is a multisensory perception. The fact that both are additive present could therefore explain the improved performance of subjects with osseointegration. Therefore, exploring the use of osseoperception as a sensory feedback mechanism to improve the sense of ownership of the prosthesis seems feasible [Clemente *et al.*, 2017] [Li and Brånemark, 2017] [Bhatnagar *et al.*, 2015]. Osseoperception combines sensations from mechanical stimulation's in the range of 25 to 650 Hz (maximum sensitivity at 250-350 Hz) and sound induced sensations due to bone conduction into the cochlear which can be perceived in the range of 0.1-10 kHz (maximum sensitivity within the 2-5 kHz range).

A similar question concerning the static force arises in bone conduction hearing aids, where [Corliss and Koidan, 1955] reported that a static force greater than 4N is recommended. The static force, as referred to in bone conduction literature, is the force pressing the transducer against the skull whilst the dynamic force is responsible for the stimulus. In [Cortés, 2002] the skull impedance was modeled and measured as a two-port network and tested for different static forces concluding that the steady state was reached around 6N and therefore confirmed that the static force should be greater then 4-6N though it does not matter whether it is 8 or 12N.

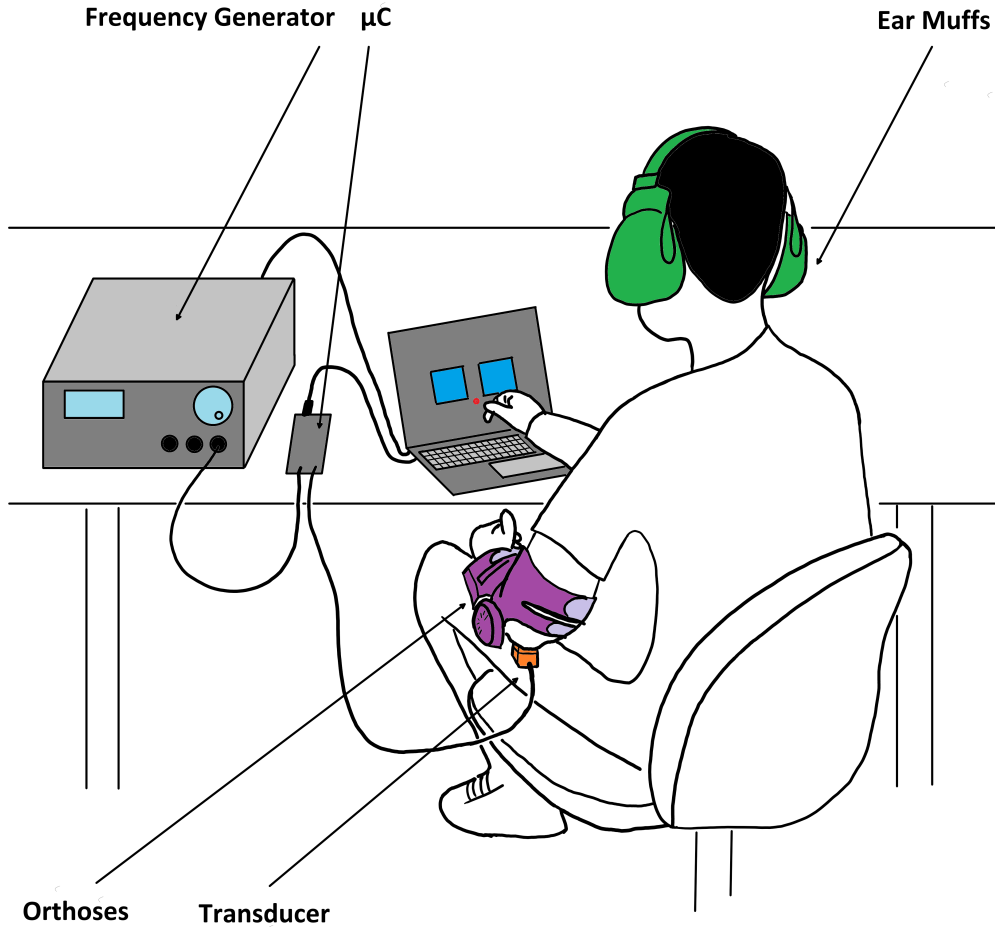


Figure 1: The test sequence is fully implemented in Matlab<sup>®</sup> and the subjects are guided through the test through a GUI explaining each step. The notebook is connected to the frequency generator. The signal is feed into a amplifier, which drives the transducer. The subject has the transducer strapped onto the left arm, making contact with the Ulna. The transducer is placed into a 3D printed PLA housing including FSR sensor. To isolate the subject from airborne sound as much as possible earplugs as well as earmuffs are used.

Force Sensitive Resistors (FSRs) allow to measure static and/or dynamic forces applied on their surface by measuring the electric resistance. They are available from several manufactures in different diameters and their advantages being cheap, little height required and capable of forces up to 10kg [Florez and Velasquez, 2010].

### 3 Material and Methods

The proposed test setup, shown in Figure 1, is built in a way to get the most flexible setup for changing stimulation as well as test setup parameters/procedures. A Matlab<sup>®</sup> GUI in conjunction with a frequency generator control the transducer and a microcontroller is used to read out the static force measured via FSR sensors. Stimulation Data is logged as well as user input using

Matlab<sup>®</sup> and the developed GUI. Due to the flexible setup, variations and extensions of this test is readily possible.

#### 3.1 Experimental Setup

Figure 1 shows the test setup consisting of a Windows surface book 2 (Intel Core i7-8, 16GB RAM, Windows 10<sup>™</sup>) as input and control unit. The frequency generator (TTi<sup>™</sup> - TG4001) is connected via an USB to RS232 converter to the notebook. The bone conduction transducer (RC-BC08A - 8Ω 300mW) is controlled using a class D stereo amplifier (maxim integrated<sup>™</sup> - MAX98306, adjusted to 6dB amplification) which is connected to the frequency generator. The used FSR sensor (Interlink Electronics<sup>™</sup> - 402 Round Short Tail) has been implemented as the upper resistor in a voltage divider with a 3.3kΩ resistor and the voltage drop

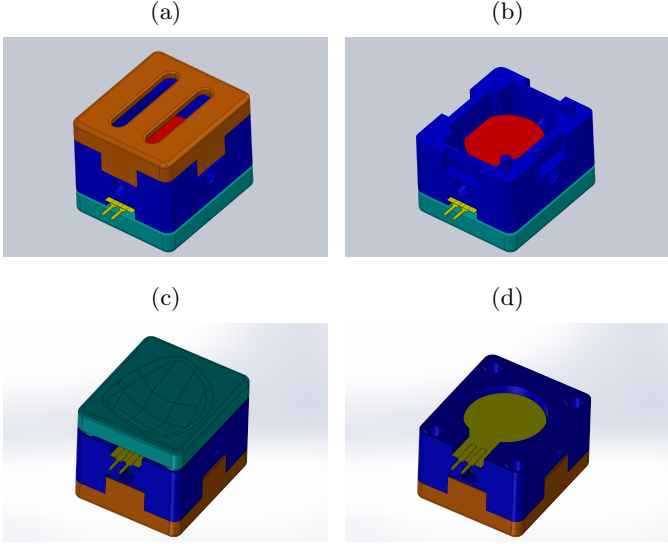


Figure 2: (a) The overall transducer unit with (b) the top cover removed to show the transducer (red). Turning the unit upside down (c) shows the bone interface (petrol blue) with the convex shape to accommodate the ulna bone shape and (d) the placement of the FSR sensor (yellow).

measured using a microcontroller (Arduino<sup>TM</sup> Mega 2540) which is connected via USB to the computer and delivers the measurement data via vCOM interface to Matlab<sup>®</sup>.

The transducer is placed in a 3D printed PLA housing, see Figure 2, where as for the design of the bone interface to the Ulna bone, a 3D bone model from *BodyParts3D*, © The Database Center for Life Science licensed under CC Attribution-Share Alike 2.1 Japan, has been used. The transducer case consists of 3 pieces, a lid, a main case where the transducer sits and the FSR sensor is attached to and a bone interface which is connected to the Ulna bone. The lid has 2 slits to feed through a velcro strap to mount the transducer to the subjects elbow. To reduce varying static forces, an orthoses set to 60° flexion was used to lock involuntary elbow movements.

The Matlab<sup>®</sup> GUI was designed to lead the users through the whole experiment. First window shows a general description of the test as well as an user input for gender, age and initials. Second window explains the test and lets the user test the two different sensations (tactile and auditory) by pressing a button. The third window is the actual test asking the subject to report the experienced sensation after each stimulation and measuring the current static force. In between

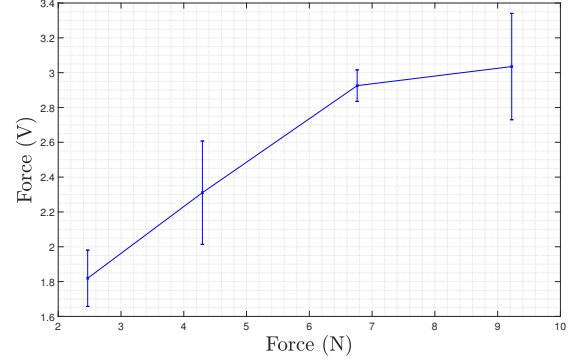


Figure 3: FSR calibration curve showing the achieved curve, using the means for further linear interpolation during the test setup. The figure shows the mean and standard deviation of 5 repetitions of four different weights [0.2517 0.4380 0.6893 0.9403]kg.

a window pops up showing the current applied static force and the target static force for the next test set in order to adjust the strap reaching the target static force.

For the sake of damping the airborne sound as much as possible the subjects were asked to put in earplugs (Moldex<sup>®</sup> Sparkplug<sup>®</sup> 29dB CL5 Uncorded Earplug) as well as wear ear muffs (Howard Leight Lightning<sup>®</sup> Hi-Visibility L3HV 33dB CL5 Headband Earmuff).

### 3.2 Calibration FSR Sensors

The FSR sensor was placed in between the transducer case and the bone interface having a cylinder with diameter of 13mm and hence covering the whole active area of the FSR sensor. The sensor calibration was achieved by placing 4 different known weights 5 times onto the FSR sensor. The mean of the achieved curve shown in Figure 3 is then used in conjunction with a linear interpolation to get the current applied force.

### 3.3 Test Procedure

The experiment is conducted with 5 able-bodied subjects (1 female, 4 male; age  $29 \pm 2.5$  years). All subjects read the plain language statement and signed the consent form approved by the Ethics Committee of the University of Melbourne (Ethics Id 1852875.1).

At the beginning of the test the subjects were seated comfortably in front of the computer. The test procedure was verbally explained and the subjects were asked to insert the earplugs and put on the earmuff on. The orthoses as well as the transducer are mounted to the left arm and the arm rested in the subjects lap. After the subjects put in their information (age, gender, initials) as well as test the sensations by pressing two

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**Algorithm 1** interval halving method

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**Require:**  $n = 1$ ,  $N = 50$ ,  $a = 0V$ ,  $b = 3V$ ,  $eps = 0.02V$ ,  $c = 0V$

```
1: while  $n < N$  do                                ▷ prevent infinity loop
2:    $s = \text{stimulate}(c)$ ;
3:   if  $s == \text{true}$  then                               ▷ Sensation
4:      $b = c$ 
5:      $c = a - (b-a)/2$ 
6:   else                                             ▷ No Sensation
7:      $a = c$ 
8:      $c = a + (b-a)/2$ 
9:   end if
10:  if  $(b - a) < eps$  then
11:    break
12:  end if
13: end while
```

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buttons before launching the test. Before starting the test, the static force was adjusted to the first of the 3 forces tested during the whole experiment ranging from  $[3\ 5\ 8]N \pm 2N$ . After that, the subject is asked to start when ready and the stimulation sequence starts. Each frequency  $[100\ 3000]Hz$  is presented alternating and repeated 5 times. Frequencies were chosen according to [Clemente *et al.*, 2017] to achieve tactile and auditory sensation. This set is repeated for each of the three static forces whilst before starting each set it is asked to adjust the static force to the static force within the given range. The threshold finding is done using an interval halving method, also known as binary search method, dichotomy method or bisection method [Kaw *et al.*, ]. It is a specific type of divide and conquer algorithm implemented as shown in Algorithm 1. It was chosen, as it is a faster approach compared to a sweep over the frequency and allows for a smaller resolution by still having less iterations. Each stimulation is presented for a duration of 1s. Then, the subject is asked if any sensation or none was experienced and the next stimulation amplitude is calculated based on the input of the subject.

## 4 Static Force Dependency

The achieved sensation thresholds are shown in Figure 4 showing the measurement of the sensation threshold for 2 different frequencies  $[100\ 3000]Hz$  at 3 different target forces  $[3\ 5\ 8]N$ .

The variability between the thresholds between subjects is much bigger for tactile sensation than for auditory perception. Though this might be misleading due to a non calibrated transducer and a frequency dependent vibration force level (VFL) of the transducer.

The change due to the static force compared to the inter subject variability is very small. As shown in Figure 4, Subject 2 has a more than twice as high threshold than Subject 4. The dependency of the sensation threshold due to the static force is very little. This is consistent to the test in bone conduction on the skull, described in section 2, though a significant increase in sensation threshold below 4N was expected. Not observing this increase suggests that the sensation threshold might be lower than the one on the skull.

The threshold for auditory sensations (3kHz) is smaller than for tactile sensations (100Hz) which could be caused due to a non calibrated transducer.

In the results there is still a slight dependency on the static force visible though the slope is small for tactile sensation and very small for auditory sensations. The same can be seen in [Cortés, 2002] where the change for higher frequencies due to changing static forces is smaller than for lower frequencies.

In [Clemente *et al.*, 2017] it was shown in experiment 1 (perception threshold) that tactile sensation has a lower threshold than auditory. At first glance, Figure 4 does not reflect this fact. Looking at VFL curves from calibrated transducers, like the B81 from Radioear, the vibration force level is 90-93dB at 100Hz (interpolated) and 100-103dB at 3kHz. Hence the 10-13dB difference, a factor of 3.16 - 4.46, incorporated into Figure 4 would change this impression.

## 5 Conclusion

This paper shows the dependency of the sensation threshold of a novel feedback interface, based on bone conduction transducers, on the static force. Carried out in the tactile range and the auditory sensation range, the experiment provides a first impression of the working principle of such a novel interface as well as the design requirements for its inclusion into a prosthesis socket.

The experiment measured the bone conduction sensation threshold of subjects given a range of static force applied by the transducer on the ulnar bone. The results show that intersubject variability is significantly higher than the within subject variations for the sensation threshold. Therefore, including the transducers into a prosthesis socket, the design needs to assure a static force bigger than 6N for all volume fluctuations. Also it suggests for future developments of the proposed feedback system to incorporate some kind of person to person calibration of the gain to account for the inter subject variability.

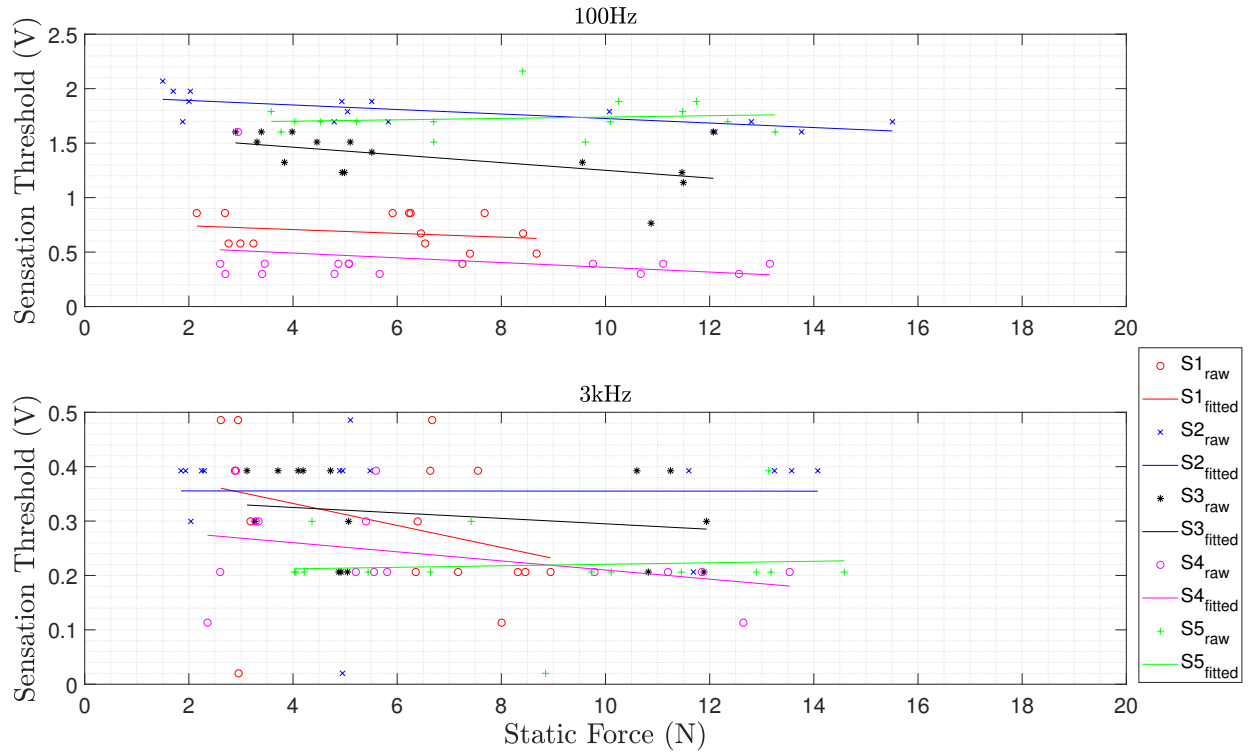


Figure 4: Sensation threshold vs applied static force. Three different static forces are targeted [3 5 8]N. The applied force is measured during the experience. Each static force target is repeated 5 times for 2 different frequencies [100 3000]Hz. In total 5 subjects are tested. Data with measured static forces outside the calibration, due to movement of the subject, is removed from the linear curve fitting.

The use of an FSR sensors, see Figure 3, in combination with fixing the transducers via straps to the elbow introduces a variation of the measured static force. Therefore, for future measurements on the static threshold a more reliable force measurement sensor having a larger force measurement range as well as a more rigid mount of the transducer is recommended. Also the small form factor needs to be addressed to include the sensor in future designs.

Furthermore, the use of a calibrated bone conduction transducer, and therefore the ability to calculate the applied stimulation force will make comparisons between different frequencies possible. To calibrate a bone conduction transducer, for the application on an elbow, similar techniques, as established for bone conduction transducers onto the skull using an artificial mastoid to calibrate should be investigate, as done in [Clemente *et al.*, 2017].

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