2021 43rd Annual International Conference of the
IEEE Engineering in Medicine & Biology Society (EMBC)
Oct 31 - Nov 4, 2021. Virtual Conference

2021 43rd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC) | 978-1-7281-1179-7/21/$31.00 ©2021 IEEE | DOI: 10.1109/EMBC46164.2021.9629539

Sensory Substitution for Tactile Feedback in Upper Limb Prostheses

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Abstract— **Present commercially available prosthetic devices fall short when it comes to providing users with accurate and non-invasive tactile feedback from their artificial limb, leading to more difficult control and leaving many at a heightened risk of device rejection. Current methods of simulating hand sensation in patients affected by upper limb loss are either invasive and expensive, or otherwise sub-optimal in their feedback mechanism. Here we propose, build, and implement a novel device for tactile feedback in upper limb prostheses. The device consists of an adaptable tactile sensing glove that can be applied to existing artificial limbs and an audio feedback system that leverages the plasticity of the brain to communicate touch to the user through sensory substitution. This device aims to take advantage of the existing pathways between auditory and tactile sensory regions in the brain by mapping force magnitude and location from the integrated force sensors on the gloves to specific volume and frequency, respectively. The device was successfully manufactured for proof of concept, and further testing with prosthetic users will aim to assess the efficacy of the device and identify potential modifications for use in research and commercialization.**

I. Introduction

An estimated 3 million people worldwide are affected by upper limb loss, but despite this large patient population, current users of myoelectric prostheses have multiple needs not fully met by the current technology. These unmet needs have led to prosthesis rejection in up to 40 percent of users [1] [2]. These needs fall into two broad categories:

1) Better prosthetic device control;
2) Mechanisms to encourage embodiment.

To the first need, current upper arm prostheses, specifically myoelectric and Brain-Computer interface (BCI)-controlled devices, are difficult for the user to control. This is partly due to the lack of feedback from the device aside from visual cues. Surveys conducted through Heidelberg University Orthopedic Hospital have shown that the most desired addition to prostheses for people with upper limb amputations is force feedback [3]. This often results in amputees either not using their devices or opting to use simpler, easier to control body-powered devices. These devices also have the benefit of providing sensation to the actuating part of the body. However, these simpler devices do not offer the same level of fidelity of movement promised by electrical prostheses, especially as this technology continues to progress.

Current commercial systems that provide closed-loop feedback for prosthetic device users have many shortcomings. Some systems use invasive techniques that involve the surgical implantation of electrodes into the residual nerve endings of the amputated arm. These techniques require residual nerve endings to be present, which is not always the case, and require additional medical procedures, making this form of force feedback undesirable for users. With non-invasive techniques, the ability to deliver feedback would be extended to all upper limb prosthetic users who either do not have the residual nerve endings required to make use of an electrode stimulating prosthetic system for closed loop feedback, or users who cannot afford a costly invasive prosthetic [1] [3].

Some potential non-invasive haptic closed-loop methods make use of surface level nerve stimulation using adhesive electrodes, but this still falls prey to the issues surrounding the availability of those residual nerves. Other systems have circumvented this problem by placing the electrodes on remote parts of the body, but this does not provide intuitive feedback to the user [4]. Some proposed devices utilize vibrational stimulus to the user [5]; however, this can cause distractions and a delayed response time. Additionally, this method is limited by the lack of intuitive relationship between forces applied to one part of the body and a vibration felt remotely from that location. This lack of intuition requires frequent and prolonged training in order for the user to adapt to the sensory substitution.

To the second need, many upper limb amputees have issues with prosthesis embodiment, where they associate or feel the device as a part of their person as opposed to something disjoint and separate. Aside from contributing to the difficulty of control, this lack of embodiment also often results in the user suffering from phantom pains and prosthesis rejection. This can be alleviated by successful integration of somatosensation [3], allowing for the prosthetic limb to be incorporated into the body image of the user. However, the method of somatosensory integration of the device is important in this process.

In addition to the need for better mechanisms of embodiment and improved means of prosthetic control, affordability of such devices is a significant point of concern. The cost for the most basic functional prostheses can range from $10,000 USD to $100,000 USD [6] for more advanced myoelectric technologies. These costs coupled with an additional expense of purchasing a costly tactile sensing device can quickly become a deterrent for the user, especially in the case of pediatric patients. Younger upper limb prosthetic users will continuously have to upgrade their prostheses to the appropriate sizes as they grow, meaning they would also need to replace the tactile sensing system accordingly, making

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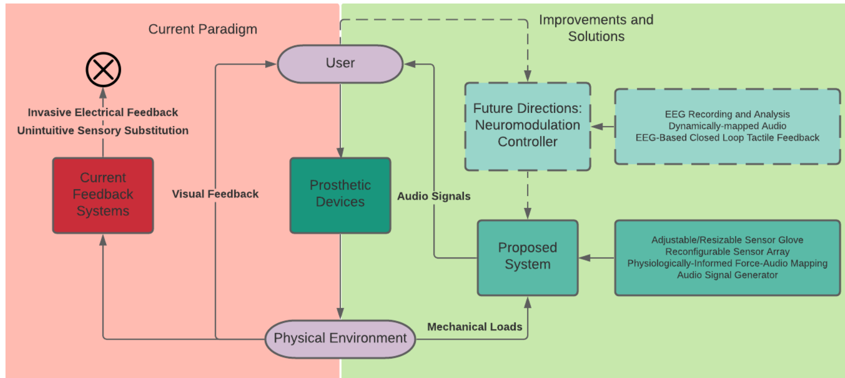

Fig. 1. Challenges and opportunities in closed-loop tactile-haptic neural interfaces, showing the current, inadequate state of sensory feedback (left) and the proposed solution (right). Dashed components represent prospective expansions of the current design to a larger neuromodulation-base system.

such additional purchases impractical and nonviable options of tactile feedback. Therefore, it is imperative to provide an affordable and adjustable means of sensory feedback for upper limb prosthetic users that can be modified to fit various sizes of prostheses.

Furthermore, introducing sensory substitution feedback devices to pediatric patients can be especially advantageous since they are shown to have higher neural plasticity than adults. Neural plasticity of the brain is defined as the brain's capability to modify synaptic connections and reorganize brain networks in response to environmental demands [7]. Thus, the rate of learning in an individual has been correlated with the development of associated brain regions which gives way to windows of increased sensitivity to environmental inputs and hence increased plasticity of the brain [7]. This results in new sensory convergences and the ability to substitute sensations with one another through proper training. However, it is important to note that neural plasticity of the brain extends into adulthood, albeit not being as robust in learning certain associations and functions as it would in a developing adolescent brain. Consequently, sensory substitution seems to be the best approach in providing haptic feedback for all upper limb prosthetic users.

Many recent studies suggest that individuals are able to quickly form relationships between audio and tactile sensations due to the preexisting anatomical integration of touch and sound in the brain. Physiologically, audio and touch are reliant on the mechanical displacement of specific receptors which are then translated into neural signals. In this way, it is evident that there is significant overlap between the basic function and structure of receptor organs in these two sensory systems [8].

From a neuroanatomical standpoint, the location and positioning of the primary auditory cortex on the superior temporal plane in the cerebral cortex further suggest ease of integration of sound and touch data as it is located adjacent to secondary somatosensory regions in the parietal operculum [8]. It is suggested that cortical networks between the auditory and somatosensory cortex, in attempts of minimizing long distance cortico-cortical connections and maximizing efficiency in their communication, can be a notable contributing factor of the shared pathways between audio and touch [8].

In light of the aforementioned information, our design criteria for the upper limb tactile sensory feedback mechanism was carefully established to consider and address all the specific patient needs, as well as to leverage feedback techniques that promise a more effective and intuitive form of sensory substitution. The current and proposed technological landscapes are summarized in Figure 1.

II. Design Solutions

To address the many needs of these patients, the following design objectives were laid out:

1) Develop a flexible and adjustable sensing interface that can accommodate a variety of prosthetic devices.
2) Develop an adjustable sensor platform that can accommodate different numbers of sensors.
3) Develop a force-to-audio mapping that relays tactile information to the user.

To achieve these objectives, a re-configurable sensing glove was developed. This glove is made of elastic fabric so that it can conform to a variety of prosthetic shapes and sizes, and can be configured with different numbers of finger sleeves based on the needs of the user. The components of the glove are fitted with force sensors that relay positional force information to a microcontroller, which in turn generates an audio stimulus to relay magnitude and location information to the user. The device consists of two primary subsystems: a re-configurable mechanical glove, and a sensor reading and audio generation system.

A. Re-configurable Mechanical Glove

Today's prosthetic market is comprised of a wide variety of device shapes and sizes, making it essential for a widely marketable device to allow for versatility of fit and configuration in order to accommodate a vast user base. The re-configurable mechanical glove functions most importantly to provide a platform for adhesion of sensors that are required for audio feedback by the rest of the system. Accordingly, a goal of the glove design was to allow for a variable number of "fingers" with the use of attachable sleeves, each sleeve capable of accompanying up to 3 sensors for a surplus of different sensor configurations. These finger sleeves themselves should also be of adjustable length to allow for optimal fit on a variety of prosthetic devices. Grounding and support for the sleeves is accomplished through adjustable straps and bands which attach to the center of the prosthetic and at the wrist. The final glove design is therefore able to be applied to a wide variety of prosthetic hands, with finger and sensor configurations that can be adjusted to accommodate the degree of feedback complexity desired by the user.

To grant modular capabilities and introduce ease of manufacturing and prototyping, materials selected for the glove are accessible and of simple fundamental properties. The integration of elastic banding allows for stretching, velcro strips ensure flexibility in attachment, and finally leather provides overall stability to the glove. Construction of the glove and introduction of materials was performed in a manner as to prevent hindering the function of prosthetic devices.

B. Sensor Reading and Audio Generation System

The sensor and audio generation subsystem consists of three major components: (1) a multiplexing sensor amplifier that can accommodate different numbers of sensors, (2) a microcontroller unit to read and analyze the sensor readings and generate the corresponding audio waveform, and (3) an audio amplifier that allows the audio stimulus to be played through headphones. The full circuit schematic can be found in Figure 2.

This system makes use of Flexiforce load sensors and a multiplexed non-inverting operational amplifier circuit. These sensors are linear force-conductance sensors, and for ease of calibration and force-audio mapping, it was desirous to maintain this linearity in the output signal of the sensor. To achieve this, the sensor was placed in the bottom leg of the op amp feedback loop. Potentiometers were used to allow for both offset and gain adjustments to accommodate variability between the sensors. This circuit is governed by the following transfer function:

$$V_0(j\omega, F) = \frac{V_{cc}\tilde{R}_1}{10^4 + \tilde{R}_1 + (330\tilde{R}_1 \times 10^{-5})j\omega)(1 + (330 \times 10^{-6})j\omega)} + \frac{V_{cc}\tilde{R}_1\tilde{R}_2}{10^4 + \tilde{R}_1 + (330\tilde{R}_1 \times 10^{-5})j\omega)(1 + (330 \times 10^{-6})j\omega)(1 + (47\tilde{R}_2 \times 10^{-12})j\omega)} Y(F)$$

where $\tilde{R}_1$ and $\tilde{R}_2$ are the values of the offset potentiometer and the feedback potentiometer respectively. These potentiometers take values between $[0, 1]k\Omega$ and $[50, 150]k\Omega$ respectively, and they work together to set the range of forces that the system can measure (and in doing so, set the sensitivity of the system). This range is set to ensure that the system amplifier saturates at a force value equal to $1.2F_{max}$, where F_{max} is the largest force that needs to be perceived. For this system and the initial prototype, this was chosen to be 15 pounds based on the range of forces that the hand applies with handling objects and on the range of forces that common myoelectric prostheses are able to generate [9]. The admittance $Y(F)$ is the admittance of the force sensor and is a linear function of force. This transfer function maintains the linearity with respect to force from the sensor through to the output voltage. At the low frequencies that are expected in this system, the transfer function reduces to:

$$V_0(F) = \frac{V_{cc}\tilde{R}_1}{10^4 + \tilde{R}_1} + \frac{V_{cc}\tilde{R}_1\tilde{R}_2}{10^4 + \tilde{R}_1} Y(F) \, .$$

The additional filtering that is incorporated into the system allows for electrical noise in the system to be reduced. It also prevents large force spikes from producing very loud audio signals, adding a level of safety for the user.

To allow the single amplifier circuit to interface with an indeterminate number of sensors, an analog multiplexer was used to sequentially connect sensors to ground. This allows for all of the sensors to be connected to the same node of the amplifier but for only one of them to be grounded, and therefore completing the feedback circuit, at a time. This allows for the number of parts and the time to calibrate the system to be drastically reduced. In addition, this system places a lower duty cycle on the sensors, limiting electrical wear. The only portion of the transfer function that would change would be the particular form of the function $Y(F)$ which would be unique for each sensor that is connected. For the current device, a 4-to-1 multiplexer was used, but a larger multiplexer could be used to incorporate a larger number of sensors.

The main system processor is a Teensy 3.6 microcontroller, an ARM Cortex M4-based microcontroller development board. This board was chosen in part because of the low cost and high clock speed (180 MHz) which allows us to add onto the system without the need for major reprogramming or redesign (the specific aims for the additional components are discussed in later sections). The board is Arduino compatible (using the Teensyduino library), so the software is in an open source format. Finally, this board also had the benefit of having integrated digital-to-analog converters (DAC) and a graphical interface for virtual audio mixers, making the audio generation portion of the system simple. This generated audio signal was then passed to an audio amplifier circuit based on the LM386 low power audio amplifier IC which in turn drove the headphones that were used to relay the audio signal to the user.

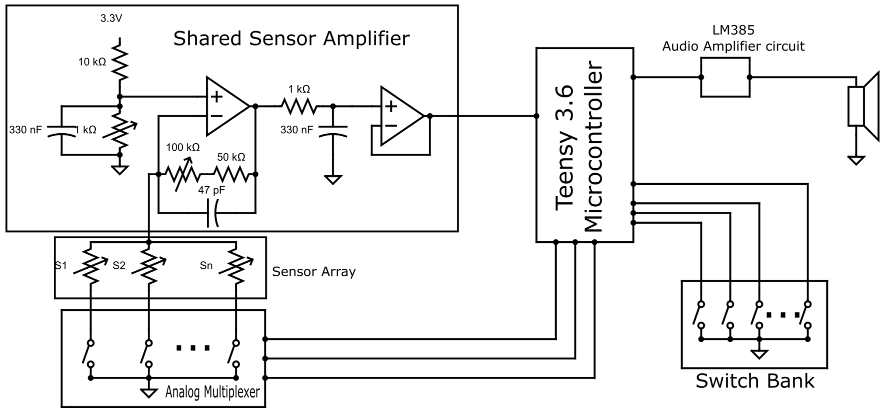

Fig. 2. Sensor reading and audio generation system schematic. Here the full electrical system schematic is shown. The shared sensor amplifier allows for tuning of the system sensitivity with a limited number of components. The sensor array, connected to ground through an analog multiplexer, can be expanded or reduced to meet the need of the user. The Teensy 3.6 microcontroller serves as the central processor of the system and generates the audio signal fed to the LM386 audio amplifier and speaker.

C. Software

The control software for the system was written using the Teensyduino library and the Teensy Audio System Design Tool. The control software includes both calibration protocols (as outlined in a future section) and standard use protocols which perform the force to audio mapping. During this standard use protocol, the system cycles through the connected sensors, and for each sensor, 10 readings are taken and averaged. Using the calibration equations that are saved for the sensor, this averaged reading is then converted to a true force reading, which is then passed to the audio generator to produce the output signal. This is achieved by changing the gains on various virtual mixer channels.

D. Force-Audio Mapping

As previously alluded to, this design centers around the goal of converting force signals measured at various locations on the mechanical glove to an audio signal that will relay this information to the user. For the current version of the device, this is achieved by mapping the force location to a frequency and the force magnitude to a volume. The frequencies associated with each sensor are fixed values and were chosen to ensure that multiple individual frequencies could be identified, but this could be optimized with further patient experiments.

Two different force magnitude-audio volume mapping equations were tested. The first took advantage of the natural log-linear relationship between volume and level of perception in humans to maintain the linearity of the force sensors through the audio transmission. This was achieved using an exponential relationship:

$$V_i(F) = \frac{1}{n}\left(\frac{1}{k-1}\right)\left(e^{\frac{F \ln k}{F_{max}}} - 1\right)$$

where $V_i(F)$ is the volume associated with the i^{th} sensor, n is the total number of sensors, and k is a shape parameter that determines the steepness of the exponential curve. This mapping requires that within the working range of the sensor ($F \in [0, F_{max}]$), the volume associated with that sensor never exceeded the fraction of the total volume allocated to each of the sensors, and that $\sum_i^n V_i(F_{max}) = 1$. This is needed because the total magnitude allowed by the audio mixer is 1. The true volume of the signal can be set by adjusting a potentiometer included in the LM386 audio circuit.

The second mapping sought to exploit that the most important range of forces were relatively low compared to the maximum force the system allows. This is because the forces associated with initial contact and that are needed to lift objects are fairly low. In this mapping, a larger increase in volume per step in force is associated with the low force range, and this relationship decays as force increases. This was achieved with the following reverse exponential mapping function:

$$V_i(F) = \frac{1}{n}\left(\frac{k}{k-1}\right)\left(1 - e^{-\frac{F \ln k}{F_{max}}}\right)$$

where the parameters have the same values as in the log-linear mapping, and have the same requirements about the volumes and force values. As with the location-frequency mapping, these volume mappings are preliminary, based on human physiology and limited experimentation. The final magnitude-volume mapping function can be further optimized with patient trials.

III. Design Implementation, Prototyping, and Testing

A prototype of the modular glove and sensor system was created and calibration protocols for the sensor circuit were developed. The system was tested to ensure the reliability of both the mechanical and electrical subsystems, and basic use tests were conducted.

A. Modular Glove Prototyping

The modular glove consists of leather, velcro, and elastic banding. The prototyping begins with the elastic finger sleeves. As seen in Figure 3-a, each sleeve has a pocket

Fig. 3. (a) Dorsal view of three finger sleeves with incorporated sensor wires. (b) Central palm unit for Sensor integration (c) Ventral view of complete sensing glove. (d,e) Modeled Sensing Glove.

constructed of leather and elastic to allow for proper stability for sensor adhesion at the fingertips and stretching to fit various fingertip sizes. Each finger sleeve pocket has leather on the ventral side for sensor adherence and woven elastic material sewn horizontally in order to allow for stretching when a finger is inserted. Braided elastic is then sewn vertically to the pockets to accommodate for different finger lengths and implement a more durable material. The ventral side of each band is sewn with velcro to allow for attachment to the central hand unit seen in Figure 3-b. This palm unit consists of a large piece of leather sewn into a wrist strap and consists of horizontally placed velcro strips that coincide with the vertical strips on each finger sleeve to allow for placement of the sleeves. The central unit allows for sensor finger sleeves along all 5 fingers, but can be seen sporting 3 in Figure 3-c. In addition to velcro strips that run on the interior of each finger sleeve, there are smaller velcro squares sewn over the tops of the sleeves in order to accommodate additional sensors in the future with the implementation of velcro rings and leather patches.

When the modular glove is worn as seen in Figures 3-d and 3-e, the central unit is slipped on first with the wrist strap secured. Each finger sleeve can then be slipped onto the corresponding fingertip, pulled to the desired length, and secured onto the central unit. Each finger sleeve contains sensor wiring channels along the dorsal side to help organize the wires down to the rest of the system. This mechanical sensing system is then integrated into the electronic audio feedback system by connecting the circuitry wiring.

B. Sensor System Prototyping

The sensor system was built into a permanent perf board (Figure 4) according to the system schematic presented earlier. Additional power circuitry was added to allow the system to run on a single 9V battery pack, and decoupling capacitors were added for each of the ICs to help minimize the power rail noise. Female headers were placed at the edge of the board so that the sensors could be place remotely on the mechanical glove, and added and removed as needed either to achieve different configurations or to assist in testing.

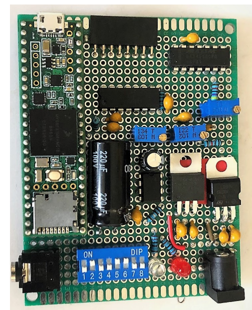

Fig. 4. Prototype Circuit System. The top portion of the circuit implements the sensor amplifier circuit, with the blue potentiometers being used to set the offset and gain. The far left of the board holds the Teensy 3.6, and the lower half of the board contains the audio and power circuitry.

C. Calibration and Testing Protocols

Calibration protocols were developed to ensure proper force reading from the sensors. Because this system shares a single sensor amplifier, this calibration needed to take place over two steps. In the first step, the potentiometers in the circuit were adjusted to ensure that all sensors are unsaturated at the F_{max}. The offset potentiometer, $\tilde{R}_1$, should be set to ensure that the reading at no load is not obscured by the noise in the signal and can be fine-tuned to achieve a specific sensitivity value. For this system, $\tilde{R}_1$ was set to give an offset value of 0.5V. Next the gain potentiometer, $\tilde{R}_2$ needs to be set, and to do this, each sensor is sequentially loaded and the force-voltage curve is observed. This was accomplished using a desktop oscilloscope instead of reading the value into the microcontroller. This gives an estimate for the sensitivity of each sensor, and then the most sensitive sensor (the one that will saturate at the lowest force value) is loaded with $1.2F_{max}$. The gain potentiometer was then set so that at $1.2F_{max}$, the amplifier was just under saturation. After this is set, the software calibration can take place. This was performed by applying no load, $1.2F_{max}$, and three intermediate loads sequentially to each sensors and having the microcontroller read and store these values. The value of load that was applied was entered to the microcontroller using the serial monitor, and using these values and the sensor reading values, a linear fit is performed for each circuit to obtain the calibration curve. The calibration parameters were then stored in non-volatile memory to ensure this calibration only needed to occur once. The R^2 value was measured to give a metric for the linearity error in the sensors.

Multiple tests were performed to characterize the sensor circuit. First, a contact trial was performed where two sensors were pushed together and the calculated force for each sensor was recorded from the microcontroller. This was to ensure that the system showed consistent force readings between sensors with different calibrations. Creep and hysteresis tests were also performed by applying five step loads to a sensor, allowing the system to sit for one minute and recording the time series data. The step values were then sequentially removed to determine the changes in force values between loading and unloading curves. To quantify the creep in the system, the time series data was fit to a simple power law function of the form $F = F_0 t^b$ where F_0 and b were the fitting parameters and t is the time in seconds. Hysteresis was quantified by the percent difference in load values during loading and unloading.

Proof of concept usability tests were performed to evaluate the ability of a user to perform simple tasks with the device in place. All tests performed with a human operator were conducted in accordance with safety guideline set by the UCSD Bioengineering Department.

IV. Results

A. Circuit Calibration and Reliability

The three sensors that were utilized in this system were independently calibrated (Figure 5), and the coefficient of determination (R^2) was calculated to quantify the linearity of the sensors. The Flexiforce sensors in combination with our amplifier system displayed a high level of linearity with an average R^2 value of 0.99. The contact trial (Figure 6) also showed good levels of consistency between the calibrated sensors.

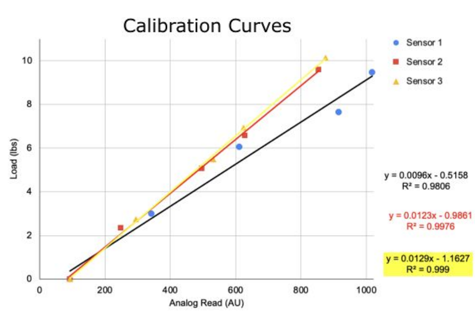

Fig. 5. Calibration curves for utilized sensors. The calibration equations and R^2 values are included for the three Flexiforce sensors that were utilized. The system was able to preserve a high level of linearity with force in the output voltage.

From the hysteresis test, it was found that there was less than 4.5 percent hysteresis across the desired range of forces. From the creep tests, it was observed that there was an inverse relationship between the force applied and the exponent parameter that was fit in the power law (at higher loads there was a lower level of drift). More thorough characterization of this creep could be incorporated in the calibration for these sensors, but this is likely unnecessary due to relatively short time scales over which this system will be used, meaning that the total level of drift will likely be unimportant.

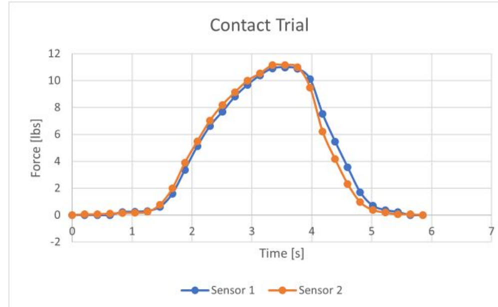

Fig. 6. Contact Trial. When two sensors are pressed together and therefore the same force is applied to both, the output shows good consistency. The slight phase shift is due to the sensors being read at different times.

B. Mechanical Device and Use Testing

As developed, the mechanical glove system was able to accommodate a variety of hand shapes and sizes, as tested on the hands of the authors. This device was not able to be tested on a prosthetic device due to facilities limitations during development, but when cycling between differently sized hands, the device was able to adjust appropriately and effectively. The modular design of the device also allowed for it to be assembled using one hand, ensuring that is it accessible to amputees. When in place the elastic material of the glove did not limit range of motion and, the finger tip sensors had sufficient friction so objects of different sizes could be lifted without slipping. This was tested with a few everyday tasks, including gripping and lifting cups, typing on a computer, and grasping small object like pencils. It was found that a particular amount of tension was required in the elastic finger sleeves to ensure that they would stay flush with the hand in the full extended position. When this tension was too low, the finger sleeves would separate from the hand and allow the system to be caught on objects.

The integrated system displayed the ability to produce compound audio stimuli that corresponded to force location and magnitude. In order to validate this functionality, a team member gripped and lifted various objects using the sensor glove whilst receiving audio stimulus from the headphones. The user reported an increase in volume especially when gripping stiffer objects relative to softer ones as well as hearing various tones depending on how many sensor fingers were used when lifting the object. These preliminary results seem promising for the tactile feedback device, however, further testing to determine the level of ease in the ability to learn using the device must be conducted. These initial tests were conducted with the log-linear force-audio mapping, and they revealed that, while this mapping followed physiological principles, it provided low volume resolution in the range of forces associated with contact, grasping, and lifting light objects. The reverse exponential mapping was then tested and it provided more useful audio information in these ranges.

V. Discussion and Conclusion

The manufactured prototype serves as functional proof of concept that the proposed system successfully translates force information measured from fingertip sensors to an audio feedback stimulus. The re-configurable modular glove design provides a structural platform for buttressing the force sensors, and is adjustable to various sizes as well as sensation complexities with the capability for additional fingers and sensors. The glove has yet to be tested on a prosthetic limb, but fits a variety of human hand sizes and thus is predicted to accomplish the goal of broad accommodation. There are several improvements that could be made to the physical glove design in addition to the incorporation of auxiliary fingers and sensors, including better containment of the sensors and their wires, and cosmetic improvements that would accompany professional manufacturing.

The amplifier circuitry and microprocessor unit as manufactured allowed for different numbers of sensors to be attached and for those sensors to be independently calibrated while still using a minimum number of components. It also effectively maintained the force linearity of sensors. The developed system provides a simple but effective platform on which to build in future iterations of this system or those similar to it. This design could also be further digitized by taking advantage of digital potentiometers that are set during the digital calibration. This would provide an added level of safety in terms of maintaining a consistent calibration needed for the patient training to remain effective.

The system currently only utilizes perpendicular force sensors which are beneficial for providing information about contact force. This alone should greatly help increase the ease of control of myoelectric prosthetic devices, but still falls far short of the vast array of human somatosensation. This system and approach could be generalized to incorporate different styles of sensors that provide different information. For example, the incorporation of soft sensors in place of the rigid sensors used here could allow for information about contact angle. Other sensors that relay information about vibrations would help provide information about textures. These systems would require more elaborate information-audio mappings beyond the scope of this work.

Using the force-audio mappings presented here, we were able to successfully relay force magnitude and location data using audio signals to a user. The reverse exponential operates mostly in the lower force ranges and is therefore the more robust of the two models, as much of everyday activity falls under this force range. The exponential mapping has the benefit of more closely aligning with human physiology.

Further testing should be implemented in order to better understand the effectiveness of user learning while using the device and provide validation for research regarding the relation between tactile sensation and audio stimulation. However, a previous study utilized a similar sensory substitution mechanisms to relay gross load data from three regions on a myoelectric prosthetic device using audio stimuli. They showed promising results in increased precision, reduced training time, and overall improved efficiency for user object gripping utilizing this method [10]. This testing will also provide information about how to further optimize the mapping process for both volume and frequency and how the device will perform in real world environments.

Finally, with some additional modifications, this device has the potential to be used with integrated neuromodulation which would allow for the transition from static functions towards a dynamic force-audio mapping approach. An early goal of this project that was postponed due to lab accessibility constraints involved implementing the current system with an in-ear EEG device developed in the Cauwenberghs Lab at UC San Diego [11]. This would expand the device to include capabilities for neuromodulation and closed-loop control of the force mapping process. In this closed loop, the mapping of force to audio would be dynamically controlled based on feedback from the EEG signal to allow for more robust and accurate transmission of force information to the user. However, this would require extensive electrophysiological studies to determine the internal cognitive mapping that occurs so that the device could properly modulate the signal.

ACKNOWLEDGMENTS

We wish to extend special thanks for the guidance and support provided by mentor Dr. Gert Cauwenberghs and instructional assistant Vasiliki Courreli. We also wish to acknowledge Gerald Stark and Erika Swanson of Ottobok for their insights early in our design process, and Dr. Bruce Wheeler for his support in this project.

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