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ORIGINAL MANUSCRIPT

Design, Development and Functionality of a Haptic Force-Matching Device for Measuring Sensory Attenuation

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Abstract

In this paper we describe the design, development and functionality of a haptic force-matching device. This device measures precise sensorimotor perception by determining a subject's ability to successfully attenuate incoming sensory signals. Sensory attenuation provides a novel method of investigating psychophysical aspects of perception and may help to formulate neurocognitive models that may account for maladaptive interoceptive processing. Several similar custom-made devices have been reported in the literature; however, a clear description of the mechanical engineering necessary to build such a device is lacking. We present, in detail, the hardware and software necessary to build such a device. Subjects ($N = 25$) were asked to match a target force on their right index finger, first by pressing directly on their finger with their other hand, then by controlling the device through an external potentiometer to control the force (indirectly) though a torque motor. In the direct condition, we observed a consistent overestimation of the force reproduced; mean force error 0.50 newtons (standard error = 0.04). In the slider condition we observed a more accurate, yet small, underestimation of reproduced force: −0.30 newtons (standard error = 0.03).

Keywords Sensory attenuation · Predictive processing · Perception · Sensorimotor · Force-matching

Introduction

The human somatosensory system is constantly inundated with sensory information from the external environment and the body's internal state. Therefore, somatosensory information that is *irrelevant*—such as that from our own movement—needs to be filtered out in order to more effectively distinguish, and attend to, the sensory information that carries greater evolutionary importance (Brown et al. 2013). One neural process by which this occurs is the prediction of sensory consequences of self-generated action (Wolpert and Flanagan 2001). An example of this occurs during human movement: internal models within the central nervous system predict the outcome of actions via an efference copy of the motor command (Blakemore et al. 2002). Thus, if the predicted sensation associated with that movement corresponds to the sensory feedback, a sense of *agency* is experienced, which is the subjective awareness of initiating, executing and controlling one's own volitional actions in the world (Brown et al. 2013). In contrast, a mismatch between prediction and sensation suggests an external event has occurred and the individual does not experience agency.

A key neural component of this process is that the predicted sensory stimuli associated with movement are compared with the actual sensory feedback, which partially cancels out sensory consequences of self-generated movement (Bays et al. 2005; Shergill et al. 2003). This process is termed perceptual sensory attenuation, which is a reduction in the perception of the afferent input of a self-produced tactile sensation due to the central cancellation of the reafferent signal by the efference copy of the motor command to produce the action (Palmer

Highlights • The force-matching paradigm provides a novel method to investigate psychophysical aspects of perception.
• We provide the mechanical engineering and software design necessary to build a force-matching device.
• Our device functions similarly to previous devices.

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et al. 2016). Sensory attenuation helps to explain the phenomenon that we perceive a reduced sensation to stimuli such as light touch, compared with when the same stimulus is applied externally (Bays et al. 2005; Shergill et al. 2003), as well as the inability of carrying out certain behaviours such as tickling ourselves (Blakemore et al. 1998).

Sensory attenuation is conceptualised within a framework known as *predictive processing*. This describes a mental process in which the brain is constantly predicting sensory signals using prior knowledge of the world, which consequently drives perception and action (Clark 2013). Within this framework, descending predictions from higher or deeper levels of neuronal hierarchies are compared with lower-level representations to form prediction errors (Seth and Friston 2016). These prediction errors, or mismatch in signal, are passed back up the hierarchy to update higher cortical representations. A fundamental aspect of this system is to minimise prediction error, and this may be achieved by weighting sensory evidence or updating prior predictions (Friston 2005). An example of this is the integration of sensory evidence with predictive signals from forward motor models which lead to sensorimotor attenuation (Wolpe et al. 2016). A failure of sensorimotor attenuation may result in false inferences about the causes of self-made acts, and this has been suggested to explain symptoms of schizophrenia (Shergill et al. 2005) and functional motor syndromes (Parees et al. 2014).

An experimental paradigm to quantitatively measure sensory attenuation is the force-matching task (Shergill et al. 2005). In this task, subjects are asked to match a force delivered to their finger, either by pressing directly on their own finger with their other hand (known as the direct condition) or by controlling the device using an external potentiometer to control the force indirectly through a torque motor (known as the slider condition). In previous research, it has been shown that healthy people consistently generate a greater force and tend to overestimate in the direct condition when compared to the slider condition (Wolpe et al. 2016; Wolpert and Flanagan 2001). The excess force produced in the direct condition reflects the sensory attenuation phenomenon, of which there is a reduction in the perception of the afferent input of a self-produced tactile sensation.

The force-matching task relies on a device which integrates haptic technology and complex electrical engineering. Custom-made devices have been designed by researchers around the world, most notably by Wolpert and colleagues, who pioneered experimental research using this task (Shergill et al. 2005; Wolpe et al. 2016). Subsequent researchers have either developed similar bespoke devices (Valles and Reed 2013; Walsh et al. 2011) or have re-commissioned existing haptic technology (Parees et al. 2014). A challenge for this research paradigm is consistency across different research groups of force delivery and measurement of force applied by participants. Specifically, this may be due to materials used, for example, the load cells which detect the reproduced force come in varying quality and sensitivity, which may be particularly relevant in older devices. In addition, computer and software capacity has significantly improved over time, which may correspond to improvements in lag, delay or any dropped frames during the experimental paradigm. These issues, combined with a lack of transparency and detailed description in the literature, has resulted in low-level outcome diversity in previous research.

While different bespoke devices may appear to operate on the same principles at a high level, as stated above, details of force delivery and measurement methods vary, or are not reported. This results in difficulties reproducing such precise instruments, where specific details are critical. The aim of this paper was to describe the design, development and functionality of a force-matching device used for experimental psychophysiological testing such that it could be reproduced by any other research group. We also aimed to present data that validate the underlying theoretical model of sensory prediction.

Material and methods

Device design and development

Figure 1 illustrates our force-matching device. The haptic system consists of a 200 W Maxon RE motor, FS20 low-force compression load cell, Bourns PTF series long-life slide potentiometer, EPOS4 50/5 positioning controller, AEDL 5810 encoder and a Raspberry Pi interface. Device functionality includes (1) impose a variable target force (1 N, 1.5 N, 2 N and 2.5 N) to a participants' right index finger, (2) external control of the motor via a potentiometer for reproduction of the target force and (3) log all output force data whilst the

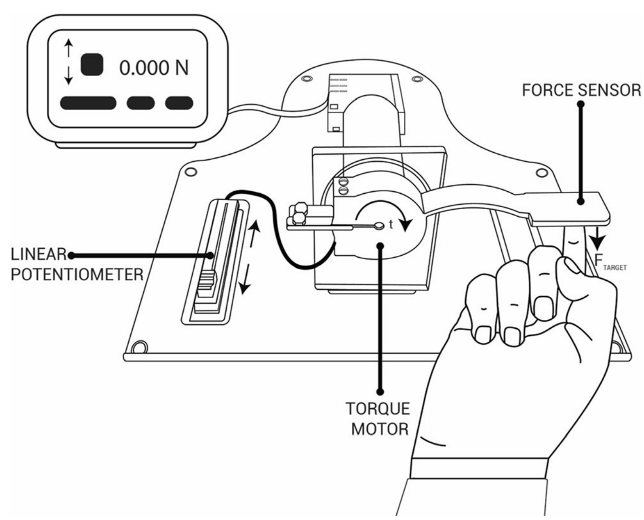

Fig. 1 Force-matching device

participant is reproducing the target force in both the direct and slider conditions.

Mechanical design

Hardware

The force-matching device is built in two main components: the haptic system and the user controller. Figure 2 shows a block diagram of the mechanical set-up. The haptic system is based on a Maxon RE50 brushed 200 W motor driven by a Maxon EPOS 50/5 controller working in position control mode. The motor includes an AEDL 5810 encoder with 5000 counts per turn over three channels with a line driver that adds a single pulse per turn for reference. The combination of the two main channels and the 5000 CPT allows for 20,000 quad counts per turn, yielding a precision of 0.018 degrees per count, as seen in Fig. 3.

The force from the motor is delivered via a bespoke-designed lever. The need for a custom design was to enable simultaneous shaft fitting, neutral balance, solid force transfer and compression cell support (see Fig. 4). The length of the lever was crucial to find the right balance between comfort of operation as well as generating enough force. With a 242 mNm/A torque constant, the chosen RE50 Maxon motor can comfortably comply with the requirements using a 15 cm-long lever, see Eq. 1.

$$F = \frac{T_k \times I}{d} \tag{1}$$

where T_k is the motor's torque constant, I is the current, d is the radial distance where the force is measured and F is the resulting force.

For continuous operations:

$$F_{cont} = \frac{0.242\ (Nm/A) \times 1.89\ (A)}{0.15\ (m)} = 3.0492$$

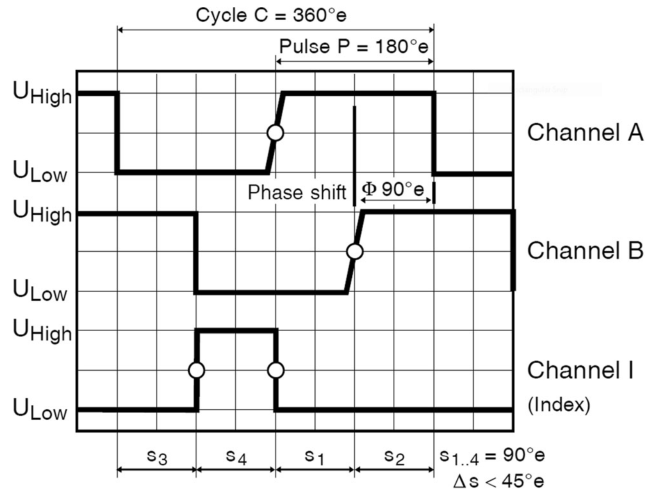

Fig. 3 The 90-degree phase shift of the two main channels allows the system to identify four different states, quadruplicating the encoder counts

For extended operations (up to 71 continuous seconds):

$$F_{max} = \frac{0.242\ (Nm/A) \times 5\ (A)}{0.15\ (m)} = 8.06\ N$$

A silicon piezoresistive compression cell was chosen for its 0–5-N sensitivity range delivered as 0.5 to 4.5 V range. The user controller is based on Raspberry Pi 3B with a quad-core ARM Cortex-A53, 64-bit running at 1.2 GHz with 1 Gb of RAM. The user input is via a 7-inch touch-enabled screen. The device is self-contained in a free-standing platform allowing the operator to freely move around the haptic device.

User interface and control algorithms

Software architecture

The goal of the software is to manage the behaviour of the haptic system and to respond to the user input based on the designed experimental protocol. It is built on four different

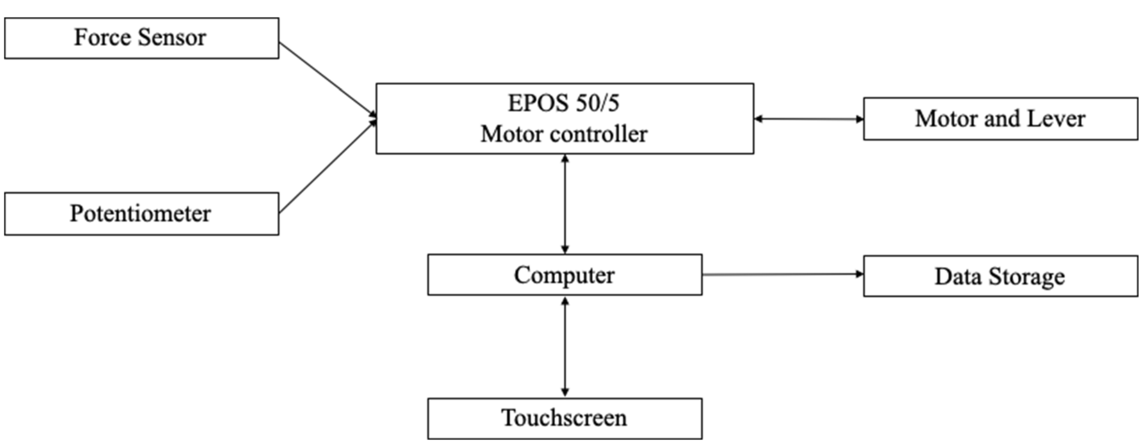

Fig. 2 Block diagram of the force-matching device

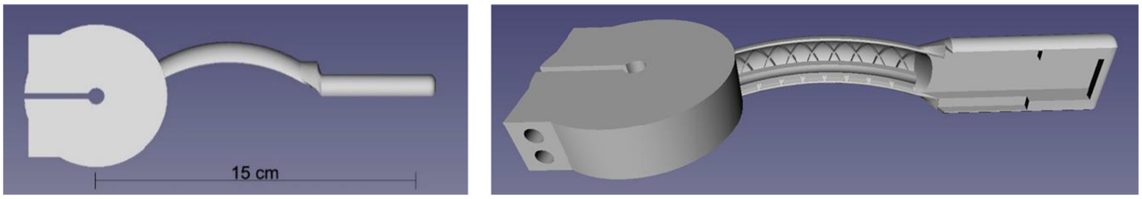

Fig. 4 The pivot point is 15 cm from the pressure point, allowing for the expected range of forces to be achieved successfully

layers: the graphical user interface (GUI), the application programming interface (API), the controller server and the controller firmware (see Fig. 5).

The GUI and API are written in Python. The controller server is written in C++. The communications between the API and the controller server are via transmission control protocol (TCP). The API sends controlling commands such as enable/disable, move up/down and go to a position. The server transmits the pressure cell and potentiometer output. The communications between the controller server and the motor controller are via a USB, using Maxon proprietary libraries. The programming code required for the device is available through the Open Science Framework. There are two files; the .py is for the GUI and top-level functions, while the .cpp is for the server and low-level functions including communication with the EPOS4. These files are available through the following link: https://osf.io/dmkcr/?view_only=25bcac9d479b4c06a7d526ee8f1f6e71

Producing and maintaining a target force

The pressure cell delivers a voltage between 0.5 and 4.5 V proportional from 0 to 5 N. The sensor has been calibrated using several weights to match 0–5 N output range. The conversion from voltage to newtons is done at the API level using the calibrated values. The pressure measured by the cell is constantly displayed by the GUI.

With the motor working in position control, when enabled, a given position is held. Pressure from the user against this position will increase the pressure measured by the cell and will be held until the motor force is overcome. When a finger is held under the pressure sensor and the lever is moved down, the force measured will increase. This is the basic mechanism of how the pressure is generated and measured.

When a target force is requested, the system will move the lever downward until the force is matched. This motion and the resulting pressure are monitored and controlled via a proportional, integral and derivative (PID) feedback algorithm. The system assumes that the finger of the user will be held in place or displaced only by a small amount. If the resisting finger is displaced leading to a large downward motion of the lever, the system will stop trying to match the force to prevent mechanical damage.

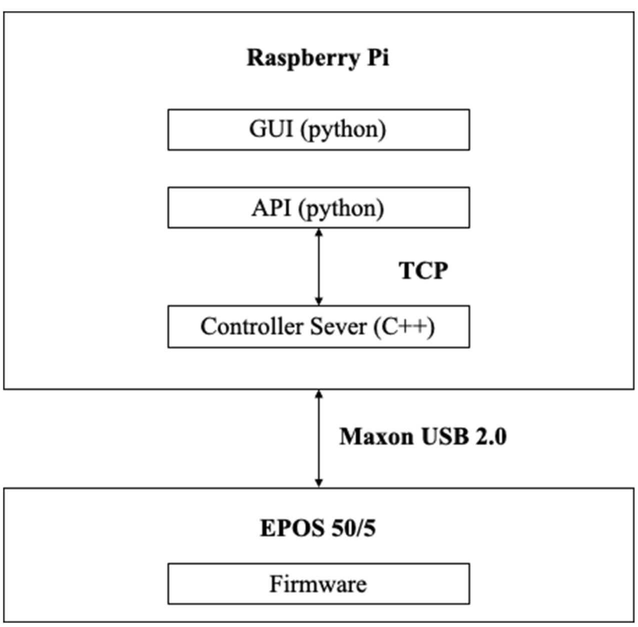

Fig. 5 The software architecture builds on the EPOS controller firmware. The layered structure allowed for a quick development time by isolating the functionality of each section

Integration of an external potentiometer

Part of the experimental protocol requires an externally driven condition. We implemented the use of a potentiometer for that task. Displacements on the potentiometer are translated into a proportional downwards motion on the lever, increasing the force exerted on the finger of the subject. The potentiometer is connected to one of the analogue inputs of the Maxon EPOS4 motor controller. The status is monitored via the USB gateway on the controller server. This information is, in turn, relayed via TCP to the API to be used as an offset from the home position.

The potentiometer outputs resistance in a logarithmic scale as a function of distance travelled. This effect created unwanted and unintuitive behaviour when trying to match the desired force. To address this issue, a linearising function was added to convert the potentiometer's output to the corresponding change of position of the lever. In order to define the linearising function, potentiometer output data were collected at equal space intervals. The relation between distance travelled on the potentiometer and voltage output were plotted in linear regression software. The resulting coefficients of the fourth-degree polynomial are the basis of the linearising function. This function is executed on the API level.

Data output

The goal of the data output is to log the force measured and a time stamp. Data is written in a CSV format (time, pressure), one line per datum. This structure is kept simple to simplify the data-reduction options later in the process.

Only two tools are given to the operator to control the data output. One is the capacity to restart the timer, and the second is the ability to start logging into a new file. Files are named with the structure Data# (where # represents a sequential number). Every time a new file is requested, an internal function checks for file names of the defined format. When it finds a missing file, it simply creates a new one and makes it available for logging experimental data.

All files are saved on an external thumb drive that is mounted automatically on the system when inserted. The GUI indicates if there is a problem with the media to warn the operator that no data is being recorded and allows for the safe ejection of the device when needed.

Display and user interaction

The GUI was designed to minimise the need for an external input and produce a clear display of the information produced (see Fig. 6). The main functions can be divided into four groups: the system set-up, the data display, the logging control and the lever control. A custom-designed image containing the control panel was used as a background. Individual areas of the image were identified to trigger actions or display information. This was achieved using the TKinter module of the Python programming language that provides specific tools for the user experience (UX) management. In addition to the user interaction tools, further logic needed to be implemented. Custom functions were developed for the timing, logging and overall control of the available options at any given time.

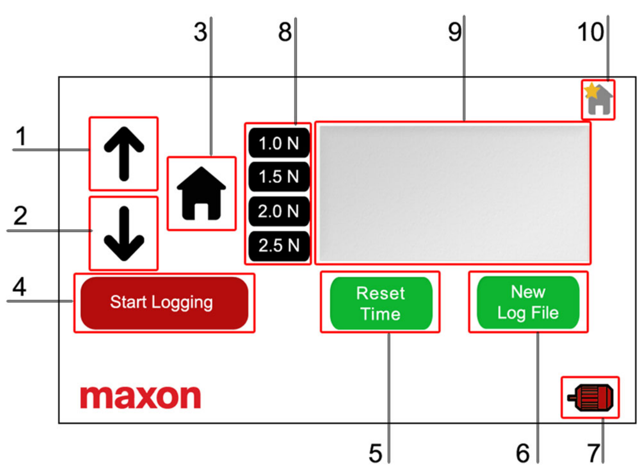

Fig. 6 The GUI is based on a custom-designed image. Functions 1, 2, 3 and 8 control the position of the lever. Logging is managed by buttons 4, 5 and 6. Button 7 enables the motor. Data is displayed on region 9, and button 10 defines a new home position in the current location of the lever

This last set of functions were essential at preventing the misuse of the device. The graphical layout of the display included buttons to move the lever up, down or to its home, user-defined position. Data management was possible via timing and logging controls. Shortcuts were provided to the four required forces. Finally, an enable/disable button will allow the user to switch the force-producing motor on or off (see Fig. 6). This button and the overall nominal force output address any safety concerns of the devious.

Device functionality

Protocol

Twenty-five right-handed participants were invited to take part in the experimental protocol in order to test the force-matching device. The study was approved by the Macquarie University Human Sciences Ethics Subcommittee (approval number: 52019574612789).

Each participant was tested in a single experimental session, consisting of two conditions (see Fig. 7): (a) the direct condition, in which participants had to match a target force by pressing directly on top of the lever, mechanically transmitting the force to the right dominant finger; and (b) the slider condition, in which participants matched the force using their opposite fingertip by moving a slider (potentiometer), which controls the torque motor. A force sensor at the end of the lever measured both the target and matched forces applied to the right finger. The comparison with the slider condition allowed for an evaluation of sensory biases in the force-matching procedure as well as an estimation of variability in basic tactile perception. Each participant was asked to reproduce four separate forces (1, 1.5, 2 and 2.5 N) on eight separate trials in a randomised order under both direct and slider conditions. The order of condition was counterbalanced across participants. The variables measured on each trial were the mean error (matched force minus target force) and ratio (matched force divided by the target force) of the force that the participant perceived compared with the target force that they actually received.

Participants sat in front of the table and placed their right index finger, just superior to the distal interphalangeal joint, underneath the force sensor which had a small silicon stopper attached to it, reducing the impact of plumpness of the finger. An ergonomic wrist and forearm support system was used to improve comfort in maintaining sustained wrist supination and to avoid any unwanted arm or finger movement (a photo of the experimental set-up has been included in the supplementary material). In the direct condition, the device exerted one of the four constant target forces in each trial for 3 seconds. After 2 seconds of rest, an auditory "go" signal instructed the participants to start matching the target force by directly pressing with their left index finger for 4 seconds

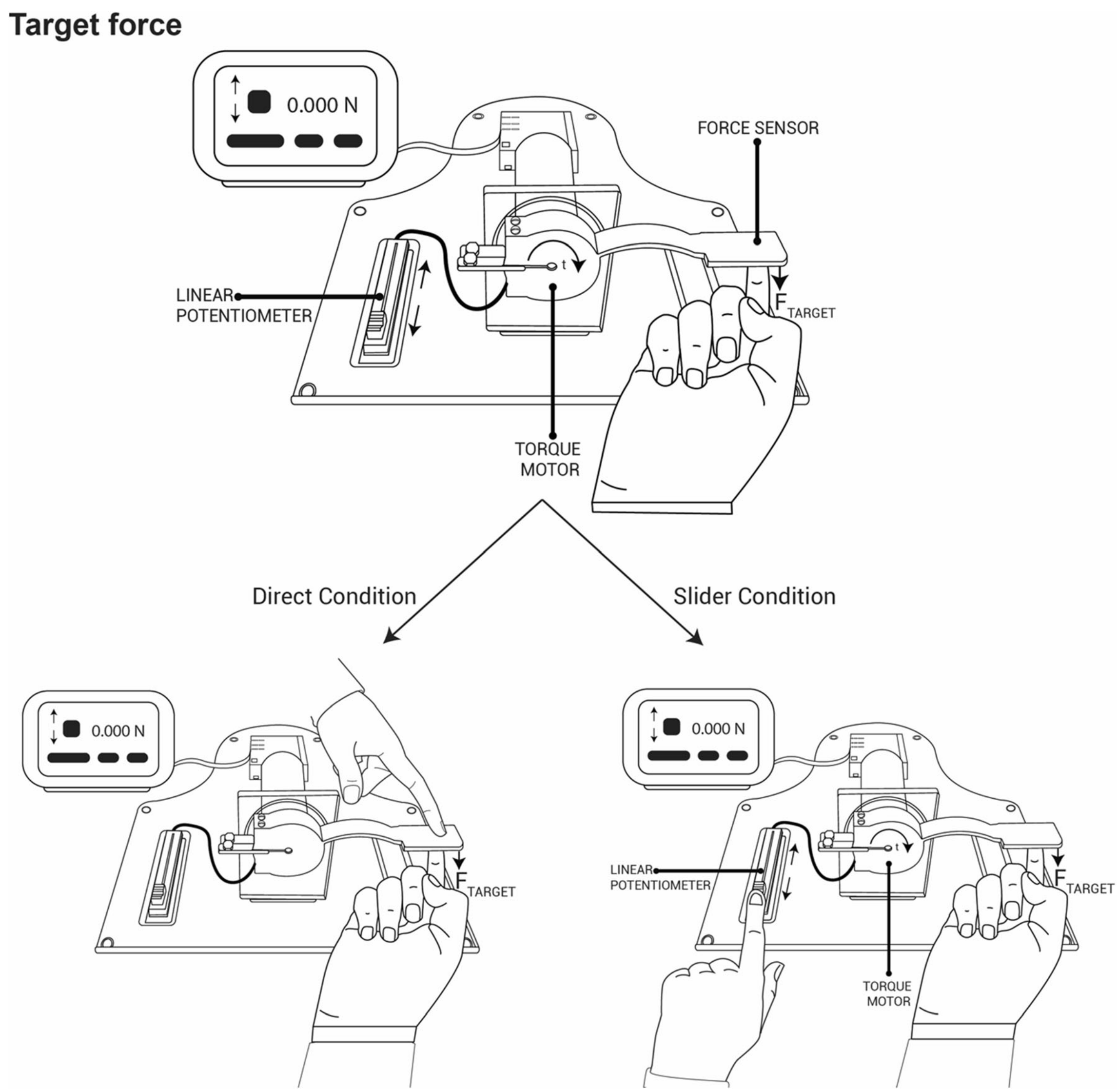

Fig. 7 Functionality of the force-matching device: A target force (1, 1.5, 2 or 2.5 N) was initially presented to the subject. In the direct condition, subjects reproduced the target force by directly applying pressure using their left index finger, while in the slider condition, subjects reproduced the target force by sliding the linear potentiometer which controls the torque motor, thus applying a force onto the right index finger

onto the force transducer resting on the right index finger. This was simultaneously coupled with manually pressing the "reset time" button, which timestamped the beginning of trial and logged the output as 00.000 seconds. A "stop" auditory signal marked the end of the trial. In the slider condition, the device exerted one of the four constant target forces in each trial for 3 seconds. After 2 seconds of rest, an auditory "go" signal indicated the participant to start matching the target force by moving the external potentiometer with their left finger. This controlled the output of the torque motor that applied a force to the right index finger. The force level generated by the subject was calculated for each trial by taking the mean force recorded by the force sensor, which was between 2.5 and 3 seconds after the go signal. The time interval used to determine the matched force has been described in the literature inconsistently, with some authors using 2–2.5 seconds (Teufel et al. 2010; Voss et al. 2007; Wolpe et al. 2016) and others using 2.5–3 seconds (Palmer et al. 2016; Parees et al. 2014). We present a side-by-side comparison of the subsequent analyses using the two time intervals in the supplementary material.

Patient-reported psychological health measures

Three dimensions of psychological health of participants were evaluated in order to measure correlates of sensory attenuation: depression, anxiety and delusional ideology. Depressive symptomology was measured with the 9-item Patient Health Questionnaire (PHQ-9) (Kroenke et al. 2001). Each item on the PHQ-9 is scored from 0 to 3, with a total score ranging from 0 (no depressive symptomology) to 27 (high levels of depressive symptomology). Anxiety was measured with the 7-item Generalized Anxiety Disorder questionnaire (GAD-7).

Each item on the GAD-7 is scored from 0 to 3, with a total score ranging from 0 (no anxiety) to 21 (high levels of anxiety). Acceptable psychometric properties of the PHQ-9 (Kroenke et al. 2001) and GAD-7 (Kroenke et al. 2016) have all been well established. Delusional ideology was measured using the Delusion Inventory (Peters et al. 2004). This consisted of 21 statements in which participants had to respond using a "yes/no" binary scale. This was designed to quantify delusion-like ideas in the general population. A total score was calculated (0–21) with high scores reflecting high levels of delusional ideology.

Statistical analysis

The matched force was recorded for each target force (1, 1.5, 2, 2.5 N) for 4 seconds for each participant. The participant's mean matched force was calculated over the interval 2.5 to 3.0 seconds after the go signal. We calculated the error (participant's mean matched force minus the target force) and ratio between the mean matched force and the target force for both conditions (ratio greater than 1 indicating generation of excessive force) at each target force. While these two indices present equivalent information, this replicates previous reports and facilitates comparison with their findings. All analyses were performed using STATA version 15 (StataCorp 2017). Simple hypothesis tests were conducted using non-parametric tests due to non-normality in some measures and bootstrapping utilised for statistical inference in the mixed model analyses for the same reason.

Results

Table 1 displays subject demographics and psychological health measures. Scores on the psychological measure are reflective of the general population and appear psychologically normal on average (Kroenke et al. 2001; Peters et al. 2004; Spitzer et al. 2006).

Table 1 Subject demographics and psychological functioning

	Participants
Age (years)	30.8 (11.94)
Sex n (%)	
Female	13 (52%)
GAD-7 (mean/SD)	4.28 (5.10)
PHQ-9 (mean/SD)	4.92 (5.98)
PDI-21 (mean/SD)	3.76 (3.64)

GAD -7: Generalised Anxiety Disorder-7; PHQ-9: Patient Health Questionnaire-9;PDI-21: Peters Delusional Ideation-21

Comparison of mean force error between direct and slider conditions

Across participants, the mean force error in the direct condition was 0.50 N ($SE = 0.04$) and was statistically significantly greater than zero (mixed model; $z = 11.46$, $p < .001$) and did not vary to a statistically significant extent across target forces (mixed model; $\chi^2_1 = 4.06$, $p = .26$). Further, the ratio within the direct condition was 1.34 ($SE = 0.04$), indicting participants on average attenuate the sensory consequences of their own actions but was observed to diminish (tended towards 1.0) with increasing target force (mixed model; $\chi^2_1 = 20.16$, $p = .002$). Within the slider condition, the mean force error was smaller than −0.30 N ($SE = 0.03$; mixed model; $z = -10.60$, $p < .001$) and also became more pronounced with increasing target force (mixed model; $\chi^2_1 = 76.29$, $p < .001$). The ratio in the slider condition was 0.85 ($SE = 0.01$; mixed model; $z = 61.37$, $p < .001$), indicating a general underestimation of the target force, and became more pronounced with increasing target force (mixed model; $\chi^2_1 = 18.08$, $p = .0004$).

The overall mean difference between the direct and slider condition with respect to error was 0.80 N ($SD = 0.08$ N; mixed model; $z = -9.84$, $p < .001$), indicating that a greater force was applied in the direct condition and denoting a clear difference in sensory attenuation and perception of the target forces. The difference between direct and slider conditions did not, however, vary to a statistically significant extent by target force (mixed model; $\chi^2_3 = 2.46$, $p = .48$). The overall mean difference between the direct and slider condition with respect to ratio was 0.49 N ($SE = 0.05$ N; mixed model; $z = -9.17$, $p < .001$), indicating a greater force applied in the direct condition. The difference between direct and slider conditions did not, however, vary to a statistically significant extent by target force (mixed model; $\chi^2_3 = 3.85$, $p = .28$). Figure 8 displays standard box plots of the mean force error by condition, highlighting this difference.

Regression of mean matched force by condition

In order to determine differences in the matched force across all force levels by condition, we conducted a mixed-effect regression analysis. Figure 9 displays mean lines of best fit of the matched force of the direct and slider conditions for each target force level. An increase in the matched force was observed in the direct condition, and this relationship flattened as the force level increased ($\beta = 0.93$, $SE = 0.04$), $z = 23.47$ ($p = < .005$)). This is in comparison to the slider condition, where a more

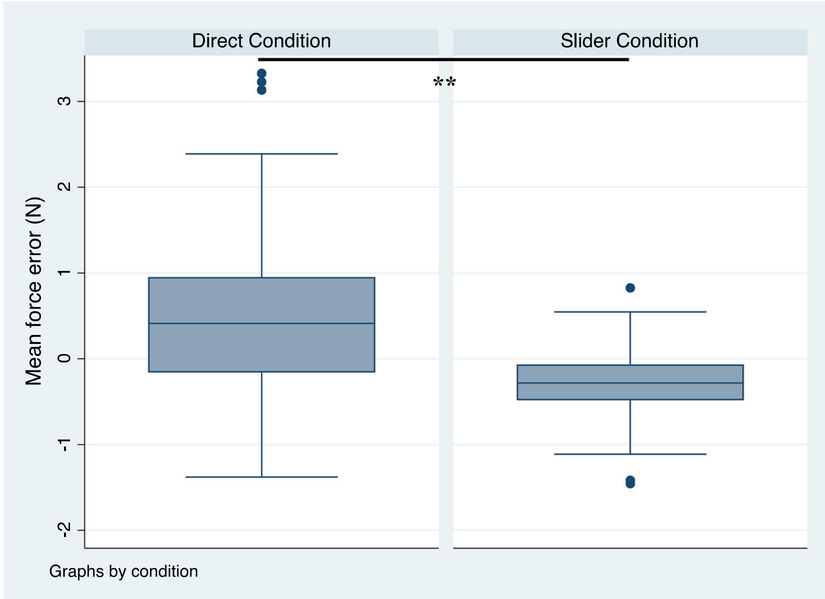

Fig. 8 Standard box plots showing the distribution of mean force error values across participants in the direct and slider conditions. Mean error was calculated as the difference between the matched force and target force across the different force levels. A positive value indicates sensory attenuation. A statistically significant difference was determined between the direct and slider conditions

pronounced flattening of the matched force at the larger force level is observed ($\beta = 0.83$, $SE = 0.02$), $z = 42.86$ ($p = < .005$). A formal comparison of the slopes was conducted using a mixed-effects multilevel regression model, with the mean matched force as the dependent variable with the target force and condition as the independent variables. A significant difference in slopes was observed ($\beta = -0.10$, $SE = 0.07$), $z = -1.47$ ($p = .14$).

Correlation of mean error with psychological variables

In order to determine whether our measure of sensory attenuation (calculated as the error; matched force minus target force) was related to psychological ill-health, we conducted Spearman's correlations. We identified a consistent non-significant relationship between sensory attenuation and

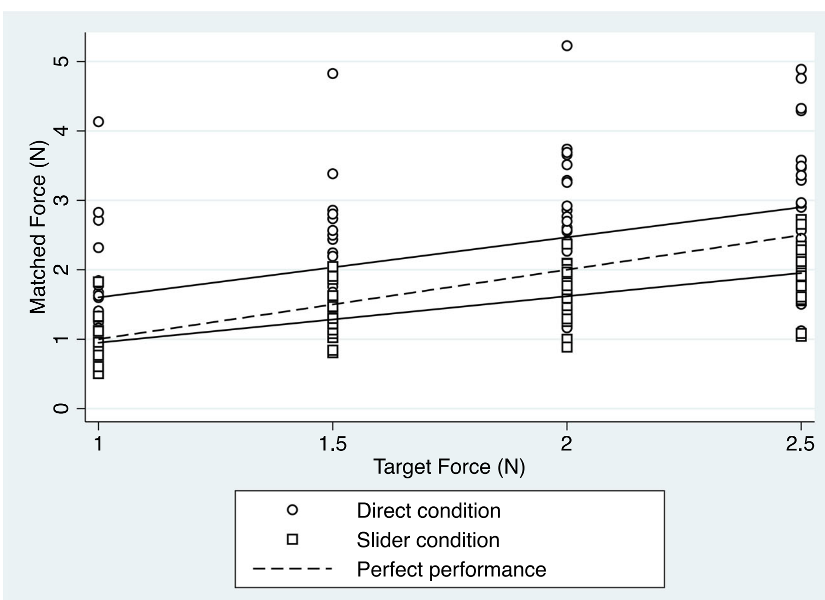

Fig. 9 Lines of best fit of the matched force against the target force for the direct (circles) and slider (squares) conditions. Reference line equates to perfect performance of the matched force

psychological variables in the direct condition: anxiety r = 0.09 (p = .67), depression r = 0.23 (p = .26) and delusional ideation r = .08 (p = 0.70). This was similarly seen in the slider condition: anxiety r = 0.13 (p = .52), depression r = .07 (p = .74) and delusional ideation r = −.16 (p = .45).

Discussion

This paper sought to comprehensively describe the design, development and functionality of a custom-made force-matching device, as well as to determine whether the model of sensory prediction was supported by the resulting data. This device was tested using a behavioural task to measure sensory attenuation to a mechanical force. Although this paradigm has been tested using a number of custom-built devices (Parees et al. 2014; Valles and Reed 2013; Walsh et al. 2011), a clear description of the mechanical engineering and associated software necessary for the construction of a force-matching device has, to date, been lacking. The detailed description provided in this paper will enable similar devices to be built by researchers in the future and ensure a consistent application of this experimental methodology.

The device functionality as presented in this paper is generally consistent with previous research (Palmer et al. 2016; Parees et al. 2014; Shergill et al. 2014; Teufel et al. 2010; Wolpe et al. 2016) and the model of sensory prediction, in which an overcompensation of the matched force in the direct condition is consistently observed. This indicates the fundamental sensorimotor attenuation phenomenon, whereby attenuation of sensation arising from one's own action depends on the integration of predictive signals and sensory feedback (Bays et al. 2006). In contrast, a more accurate estimation of the target force with the slider condition has been reported (Shergill et al. 2014; Teufel et al. 2010). It is proposed that the greater accuracy in the slider condition may be due to the reduced efference copy signals used, compared to that in the direct condition. The mechanism may include a resulting increased weighting of sensory feedback and less reliance of internal models leading to a more accurate estimation of the target force.

During testing of our device, we identified a more accurate, yet small underestimation of the perceived target force in the slider condition. We present this as a new finding, which may be explained by the finding that, in the presence of reduced predictive signals, compared that in self-generated forces, more variability in determining the target force may exist. In addition, as per our motivation of this work, differences in the materials, software and methodological processes may also lead to subtle differences in results. This point is further highlighted by comparing two different devices; in Palmer et al. (2016), the mean slider error and standard deviation were 1.18 N and 0.79 N, respectively, while in Wolpe et al. (2016), the mean error and standard deviation were 0.05 N and 0.42 N, a mean differential of 1.13 N. Notwithstanding the minor differences observed in the slider condition, the sensory attenuation phenomenon described during the force-matching task is in line with previous published results and provides a novel avenue for psychophysiology testing in sensorimotor perception.

This avenue of testing has a meaningful impact on furthering our understanding in specific patient groups who may also experience alteration in perceptual sensory attenuation. This process is an attentional phenomenon, comparable to turning up the volume or gain of a sensory channel (Friston et al. 2013). The process of attention in this context does not equate to voluntary allocation of conscious attention and is, for example, in contrast to a generalised increase in conscious body-focused attention, which is commonly seen in a number of functional or somatisation disorders (Rief and Barsky 2005). It is proposed that the increased body-focused attention identified in patients with some functional disorders may be related to an imprecise prior prediction associated with a greater weighting of sensory signals and reduced sensory attenuation (Parees et al. 2014). Much of the literature to date using this device has been limited to people with schizophrenia and functional motor symptoms (i.e. psychogenic tremor), and it is unknown whether other functional patient groups, such as those with chronic pain, may also experience alterations in perceptual sensory attenuation.

Finally, the maximum force output of the motor is 8 N and is significantly less than that described in other devices, who tested up to 40 N (Walsh et al. 2011). Our output was specifically selected to investigate lower force ranges and paired with a more sensitive load cell which had been calibrated to detect between 0 and 5 N, for as discussed by Walsh et al., the overestimation effects in the direct condition were only observed in forces up to 55% of a subject's maximum voluntary contraction. Further, it is important to note the small sample used to test the functionality of the device and related statistical analyses. Future research would benefit from a larger-scale replication of the current outcomes.

Conclusions

This paper describes the design, development and functionality of a custom-made force-matching device. The device performance is generally consistent with previously published results using this paradigm: specifically, when a passively experienced target force is matched, participants applied a larger force in the direct condition, compared with the slider condition. This device enables use of a behavioural task to investigate sensory attenuation. This study provides a clear description of the design of the force-matching device for replication by other researchers.

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Author Contributions David McNaughton: Conceptualisation: Data curation; Formal analysis; Funding acquisition; Writing—original draft

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Declarations

Data availability De-identified experimental data available on request to authors.

Consent for publication We consent to publication to *Behaviour Research Methods* and confirm that the current manuscript is not under review by or submitted to any other journal.

Code availability The programming code required for the device is available through the Open Science Framework. There are two files; the .py is for the GUI and top-level functions, while the .cpp is for the server and low-level functions including comms with the EPOS4. These files are available through this link: https://osf.io/dmkcr/?view_only=25bcac9d479b4c06a7d526ee8f1f6e71

Ethics approval The study was approved by the Macquarie University Human Sciences Ethics Subcommittee (Approval number: 52019574612789).

Conflicts of interest/Competing interests No conflict of interest or competing interest to declare.

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