FMRI and MEG Compatible Hand Motion Sensor

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INTRODUCTION

Functional magnetic resonance imaging (fMRI)) and magnetoencephalography (MEG) are two of the most important neuroscience tools, providing spatial and temporal localisation of brain activity [1]. Investigations of somatosensory function or neuroplasticity in rehabilitation greatly benefit from the correlation of neuroimaging and motion sensor data [2], particularly considering recent MEG breakthroughs which enable more accurate assessment of brain function during real world visuomotor coordination tasks [3]. However, constructing wearable sensors suitable for these environments presents a significant design challenge. Interaction between the MR environment and ferromagnetic components present safety concerns. Further electronics and changes in the static magnetic field i.e., from moving metal components, can produce interference. Existing MR compatible sensor systems, such as fibre optic sensors [2][4] or Vision-Based Motion Analysis [5] are either non-cost-effective or require line of sight from multiple angles, which may not always be feasible in fMRI or MEG rooms. We propose a MR/MEG compatible joint sensor, Fig. 1, to track finger motion during grasping tasks, common in rehabilitation or functional experiments. Using the Skinflow [6] principle of fluidic transmission and metal free construction, the sensor has the potential to track the joint angle without creating artefacts in the neuroimaging data, at a lower cost than fibre optic sensors, and without the need for line of sight.

MATERIALS AND METHODS

The sensor consists of two soft components: a soft liquid sensor positioned on the finger joint, and a computer vision (CV) system to detect the liquid level at the opposite end of the tubing. Bending of the finger joint displaces the coloured water within the sensor chamber, which is transmitted a sufficient distance from the sensitive equipment through tubing to a glass display tube. The computer vision system then detects displacement of the liquid in the display tube.

The sensor body, Fig. 1B, contains a network of fluid chambers, which are compressed during bending and displace the liquid down the attached tube. This design also reduces the resistance to the natural bending motion at the finger joints. The sensor was constructed from two moulded Sylgard 186 (Dow, Inc.,U.S.) parts glued together with SilPoxy (Smooth-On, Inc.,U.S). Silicone tubing was used to connect the sensor inlet to the CV system. The fluid level inside the tubing was obtained by segmentation of video recorded at 30 fps using a webcam (Logitech Brio). The edge within each row of pixels within the liquid segmentation was then averaged to obtain a single fluid level in mm. The total tube length from sensor to CV was 1.5 m.

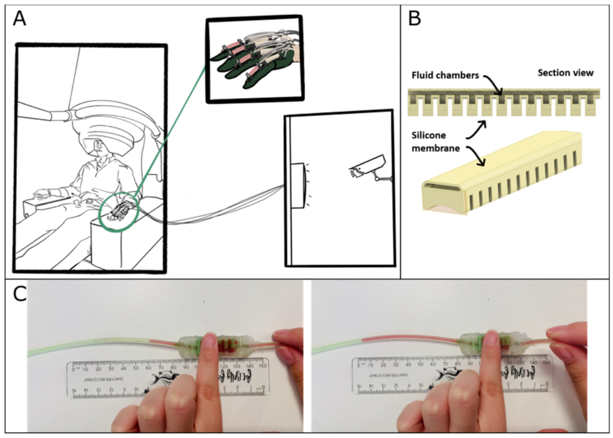

Fig 1. fMRI/MEG compatible sensor concept: (a) fluid displacement from hand motion transmitted to computer vision system away from sensitive equipment (b) sensor geometry (c) demonstration of working principle

Initially a motorised hinge was used to evaluate the relationship between bending angle and the change in displaced fluid. The sensor was connected to both leaves of the hinge, and a single leaf was driven directly by a 0.9° stepper motor (uStepper), Fig. 2A. Thus, the motor angle represented the ground truth bending angle, enabling the characterisation of the accuracy and repeatability of the sensor. The hinge joint was

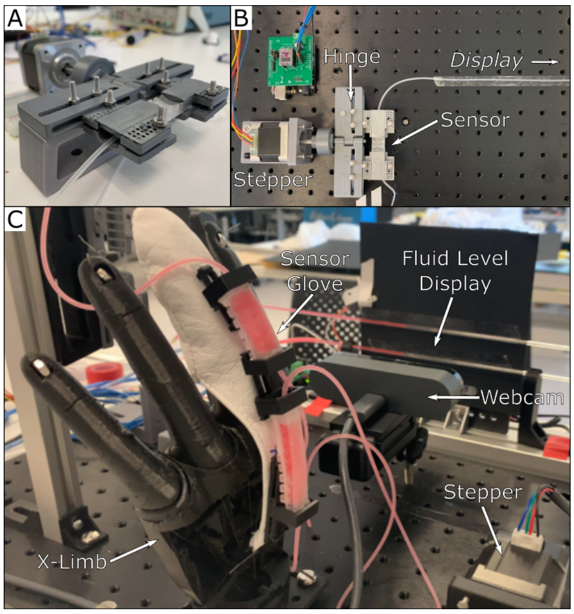

Fig 2. Experimental setup: (a) Sensor in motorised hinge (b) full hinge setup (c) sensor glove on X-Limb prosthetic

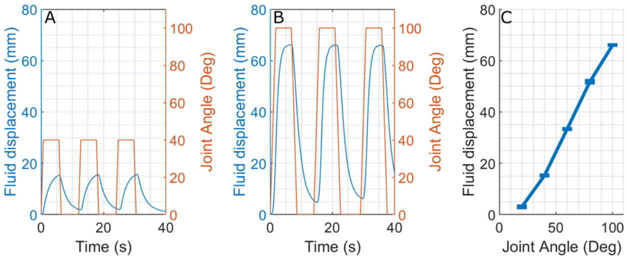

Fig 3. Hinge experiment results: fluid displacement captured by CV system during (a) 40° (b) 100° actuation (c) repeatability of final displacement with joint angle (mean±std)

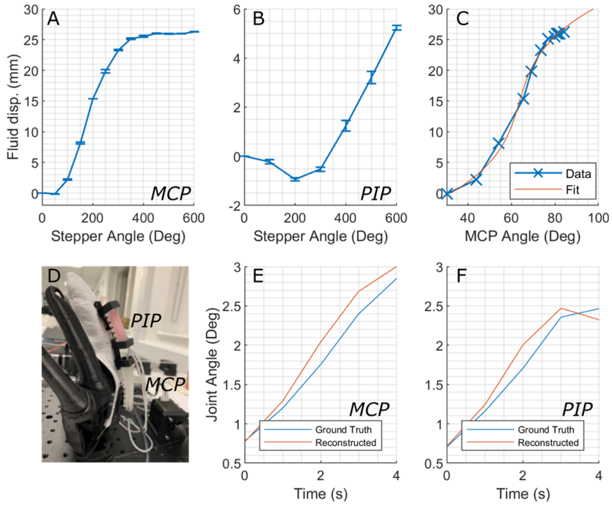

Fig 4. X-Limb experimental results: fluid displacement recorded by CV system during cable driven grip motion (a) MCP (b) PIP joint (c) comparison with ground truth angle (d) finger glove embodiment on X-Limb prosthetic (e) and (f) reconstructed joint angle for MCP and PIP

repeatedly actuated (n=10) from 20° to 100° with a 5 second settling time.

The sensor was then tested using a flexible 3D-printed cable driven finger from the open-source X-Limb prosthetic [7], Fig 2C. A repeatable grip motion was created using a stepper motor and the ground truth of each joint angle was obtained through tracking four markers using OpenCV. The sensor was placed on the Metacarpophalangeal (MCP) and Proximal Interphalangeal (PIP) joints, and data from 10 repeated grip motions was used in a polynomial regression to convert fluid level to joint angle. The joint angle of a further three repeats were used to estimate the accuracy of the reconstructed finger pose.

RESULTS

In the hinge experiments the sensor demonstrated a linear response for bending angles 20° and above (R^2=0.99), Fig. 3C, with a mean repeatability error of 0.25 mm (< 1%). The results from both the MCP and PIP joints on the X-Limb prosthetic finger, Fig. 4, were similarly repeatable with errors of 0.13 mm (< 1%). However, the correlation between fluid level measured and joint angle was not linear during the grip motion, thus a 4^{th} order polynomial was used in the regression model. The finger pose was successfully reconstructed with an accuracy of 8.8° (6.8%) with the greatest error.

DISCUSSION

Overall, the results demonstrate the principle that the finger joint angle can be sensed using fluidic transmission in a sensor with metal free construction. The sensor is also lightweight (10g per joint) and low cost, using ~ £7 of materials excluding the camera. The results of the hinge experiments demonstrate the highly repeatable and linear response of the system for angles between 20-100°. The motion of the sensors on the X-Limb finger was a combination of both joints and orientation changes, and thus a higher order polynomial fit was required to estimate the joint angle. Integrating the sensor into the wearable housing did not adversely affect the performance as the repeatability error was unchanged from hinge experiments (<1%). This was sufficient to reconstruct the finger pose during a grip motion. This demonstrates the feasibility of the sensor as a wearable device for tracking hand motion during simple grasping tasks for neuroimaging.

The dynamic response of the sensor requires improvement before it is ready to incorporate into more complicated experiments, as the settling time for the largest angle changes are approximately 4s. This lag was also the cause of the larger error in reconstructed angle when the prosthetic was in motion. This error could be reduced, as per the Hagen-Poiseuille equation, by changing the tubing material to reduce friction (e.g., PTFE) and increasing the diameter to reduce the resistance to flow. However, increasing cross sectional area would also reduce the change in fluid level and would thus be harder to detect with the current CV system. Future work in improving the CV system would also be beneficial for measuring the smaller angle changes (≤10°), which were not reliably detected.

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