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Hand-Held Microsurgical Forceps with Force-Feedback for Micromanipulation

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Abstract— **This paper presents a hand-held microsurgical forceps design with force-feedback capabilities designed for micromanipulation tasks. The device uses a customized force sensor that measures grasping forces over a range of 0-300mN and uses an actuator to exert amplified forces back on to the operator's fingertip in a mechanically-ungrounded setup. This allows perception of low force levels that are otherwise imperceptible to human touch. A customized force sensor design for the forceps grasping measurement is presented and a calibration experiment was conducted to validate its linearity and repeatability. A bench test of the device was conducted to demonstrate its intrinsic force-amplifying capabilities, with amplification factors of up to ×50 reported. A user study was conducted to confirm that the device could significantly improve human perception of grasping forces compared to conventional microsurgical forceps with the results demonstrating an order-of-magnitude improvement in force perception.**

I. INTRODUCTION

ROBOTIC systems designed for surgery aim to exploit the superior control and enhanced sensing capabilities of a robot in combination with the clinical experience and decision-making capabilities of the operating surgeon. This is commonly achieved through teleoperated master-slave systems or cooperatively-controlled robots, such as the Steady-Hand robot [1]. Cooperatively-controlled robots allow manipulation of the instrument by both the surgeon and the robot which provides steady, precise motion of the surgical tool and the ability to provide force-scaling for delicate surgical tasks. Mechanically-ungrounded, hand-held robotic surgical devices are an alternative approach to grounded systems, with the advantages of being more compact and physically unobtrusive to manipulate. Furthermore, hand-held robotic surgical devices can be integrated into the surgical workflow with greater ease.

Micron is one such hand-held device that has been primarily designed to suppress physiological tremor of the operator [2]. This is useful in micromanipulation tasks that warrant particularly precise control of the interventional tools, such as in retinal surgery [3] and cell manipulation [4]. Hand-held devices with tool-tip motion-compensation can also incorporate virtual fixtures that prevent the instrument tool-tip reaching anatomically-critical 'no-go' zones [5], [6]. More recently, these systems have incorporated force-feedback to the operating surgeon [7]. Haptic feedback ensures that the forces exerted are appropriately delicate when handling tissue. In fields such as neurosurgery and ophthalmic surgery, tissue is highly fragile, with tool interaction forces often below 7.5mN; levels that are often imperceptible to most surgeons [8]. Moreover, in the case of neurosurgery, brain tissue is frequently eloquent and has little capacity for regeneration, so the consequences of iatrogenic injury when such force thresholds are exceeded are often severe and usually permanent. The corollary is that devices that scale up tool-tissue interaction forces may reduce the risk of tissue injury, and improve patient safety [9]. Mechanically-ungrounded, hand-held devices that can perform force amplification in micromanipulation have been devised for this purpose. In a mechanically-ungrounded setup, the reaction forces generated by the device must be transmitted through the operator of the instrument. These devices utilize embedded actuators to exert forces on to the operator based on forces sensed at the tool tip. Stetten *et al.* described a force-magnifying device that provided feedback using a solenoid mounted on to the back of the operator's hand [10], [11]. Furthermore, Payne *et al.* presented a device that provided force-feedback to the operator's fingertip, forgoing a bracing or anchoring mechanism in [12].

A key requirement of force-amplifying devices is in the sensing of low force ranges below those of human perception. One such technique is the use of optical fiber sensors to provide high sensitivity force sensing in surgical instruments. Fiber Bragg grating (FBG) sensors with sub milli-newton accuracy have been integrated into surgical tools for retinal microsurgery [13] and have been implemented on the Steady-Hand robot [14]. These sensing systems have also been applied to hand-held instruments, including Micron [15] and in customized forceps [16], [17]. Micro-electromechanical (MEMS-based) sensors have also been developed for force-feedback systems [18] and in micro-gripper designs [19]. Strain gauge-based sensing has also been adopted for micro-gripping applications [20], [21]. Researchers have also devised forceps that incorporate piezoelectric sensing and actuation [22].

Intelligent force sensing at the tip of a surgical instrument allows for meaningful force-feedback whereby only the important tool-tissue forces are rendered back to the surgeon. Examples of this include the bypassing of the frictional insertion forces in both endovascular catheterization [23] and epidural needle insertion [24]. In the case of microsurgical grasping tasks, force-feedback to the operator is hindered when using conventional spring-loaded

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forceps as substantial force must be exerted simply to overcome the spring-return mechanism.

This paper presents hand-held microsurgical forceps that incorporate force-feedback to the operator based upon a direct measurement of the forceps grasping force. The system allows amplification of these forces to levels that are better perceived by the human operator in a mechanically-ungrounded configuration. A customized force sensor mounted at the forceps tip is presented with a force-sensing capability in the 0-300mN range which allows force amplification factors of up to ×50. The sensor calibration is reported and the device is validated in bench tests with respect to its intrinsic force-feedback capabilities. Additionally, a user study was conducted in order to confirm that the force-feedback can help limit the amount of force exerted in delicate grasping tasks.

II. Device Overview and Design

A. Device Overview

The force-feedback forceps are designed according to standard microsurgical forceps whereby the operating surgeon must compress a spring return lever mechanism in order to close the forceps. A force sensor is integrated into one tip of the forceps in order to allow measurements of the forceps grasping force (F_g). An actuator is then used to exert amplified forces on to the operator's fingertip. The reaction forces generated by the actuator are transmitted through the operator's grip of the force-feedback forceps. Crucially, the actuator is mounted perpendicular to the grasping direction of the forceps levers. This design feature ensures the operating surgeon can distinguish between the amplified-force feedback from the forceps tip sensor and the forces required to compress the spring levers of the forceps by exploiting the ability of the human fingertip to resolve forces spatially. A second force sensor, mounted between the actuator and the operator's fingertip allows measurements of the force exerted on the operator (F_f) which permits closed-loop force control to provide force amplification. The device is presented in Fig. 1.

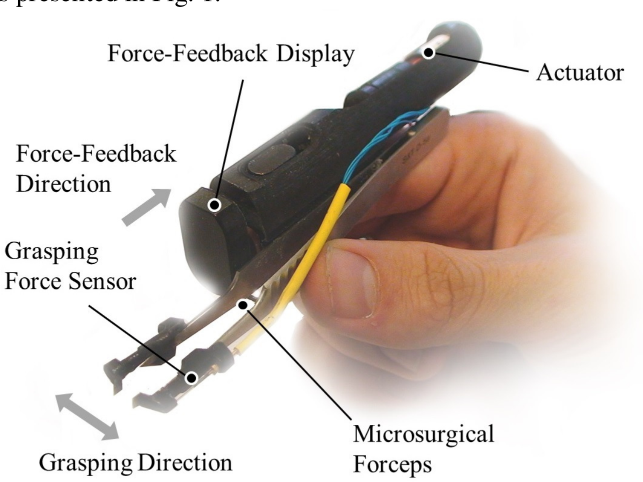

Fig. 1. A photograph of the hand-held force-feedback forceps prototype.

B. Grasping Force Sensor Design

A critical aspect of force-feedback systems for microsurgical applications is the tool-tissue force-sensing. Commercially-available, fully-integrated force-sensors are typically large and designed to measure forces at the newton-scale. With such sensors, force measurements at the milli-newton-scale require significant amplification and are therefore prone to noise interference. Another key consideration is the physical size of the sensors which must remain compact for practical clinical deployment. In consideration of these aspects, a silicon strain gauge-based force sensor (AE801, Kronex, Oakland, USA) was chosen to make measurements of the forceps grasping forces. The sensor is configured as a silicon cantilever beam which is sensitive to bending deflections. The sensor is compact (1.8mm diameter × 10mm total length) and can measure a maximum force of 120mN, leading to good sensitivity and a high signal-to-noise ratio at sub-newton force levels. The silicon strain-gauge sensor exhibits good linearity (±0.25% FSO) and low hysteresis (±0.1% FSO). Another advantage of strain-gauge sensors is their low cost and that their operation does not necessitate additional hardware, such as the integrators required for FBG sensors. A disadvantage of the AE801 is its fragility; the sensor is only rated to accept a 120% overload and is also fragile to other out-of-plane and torsional loads, making it impractical for direct deployment in a surgical device.

To address this issue, the AE801 is integrated into an assembly that utilizes an elastic beam in parallel to the sensor beam so as to support a proportion of the total grasping load. The sensor assembly is presented in Fig. 2.

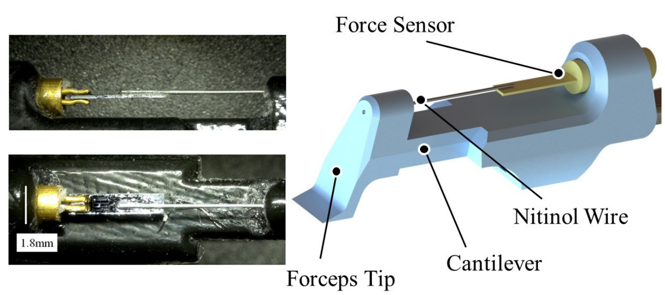

Fig. 2. Close-up photographs and illustration of the grasping force sensor assembly.

Since the AE801 is sensitive to displacements at the µm scale, the sensor is coupled to the forceps tip using a 0.17mm nitinol wire which acts as a low stiffness spring so as to prevent significant undesirable pre-loading during assembly and avoids the necessity for high precision manufacturing which cannot be achieved using the rapid prototyping methods used to manufacture the sensor assembly. The sensor was designed empirically, with the aim of achieving a minimum sensing range of 0-300mN so as to be sensitive enough at low force levels whilst remaining robust to higher loading. This was done through a process of tuning the stiffness of the parallel cantilever and the stiffness of the coupling spring through geometrical alterations to the design. The AE801 was rigidly bonded to the forceps tip assembly using a cyanoacrylate adhesive to minimize hysteresis effects.

C. Mechatronics Design and Control Scheme

A base chassis and the sensor assembly previously described are mounted to standard, stainless steel microsurgical forceps (S&T D-5A, Dumont Surgical, Gaithersburg, USA) that are commonly used for grasping tasks in micro-anastomosis. The forceps tips have a total grasping stroke of 11mm. The force-feedback forceps are designed to operate as close to conventional forceps as possible with minimal interference from the additional components. The tight spatial constraints imposed by the short, angled tips of the forceps and the grasping location of the operator's fingertip make it impractical to mount a force-feedback actuator ahead of the fingertip location. It is therefore mounted at the proximal region of the forceps, off the axis of the forceps levers so as to allow space for the operator's finger. A voice-coil actuator (VCA) (LVCM-013-013-02, Moticont) was used to provide the force-feedback. The forces generated by the VCA are then transmitted through a low friction slider mechanism which is constrained to pure axial motion through an anti-rotation keyway. The slider is coupled to a plate which makes contact with the operator's fingertip through the feedback force sensor (8438 5005, Burster). The total weight of the device is 48g. These design features are illustrated in Fig. 3

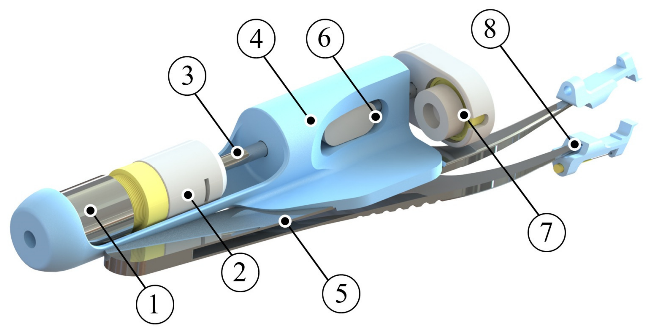

Fig. 3. Shows the key features of the device (1) VCA, (2) coupling, (3) force transmission rod (4) mounting chassis, (5) microsurgical forceps, (6) anti-rotation mechanism, (7) force-feedback display incorporating the force-feedback sensor and (8) the force-sensing forceps tip.

The addition of the mounting chassis and VCA to the elastic forceps forms a cantilever mass-spring system. Therefore, the chassis is mounted on the opposite forceps lever to that of the lever containing the force sensor. This avoids control instability arising through excitation due to the mechanical coupling of the sensor and actuator.

The control scheme for the force-feedback forceps is implemented on a multi-threaded, real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.) that operates on a 1kHz loop. Both the grasping force sensor and feedback force sensor input voltages are acquired by the controller following amplification through separate Wheatstone half-bridge arrangements before being converted to newtons by the controller. The forceps grasping force measurement F_g is then multiplied by the pre-set amplification factor before forming the set-point to a PID control loop. The fingertip force measurement F_f provides the feedback in order to facilitate closed-loop control. The output from the PID algorithm is converted to a voltage by the CompactRIO which is then received by a linear amplifier (LSC 30/2 linear 4Q Servoamplifier, Maxon) which supplies power to the VCA.

III. Experiments

A. Grasping Force Sensor Calibration

A calibration experiment was conducted to evaluate the forceps tip sensor assembly prototype with respect to its linearity and repeatability. This is important in order to validate the force-amplifying capabilities of the overall system. The sensor assembly was loaded against a force-torque (F/T) sensor (Nano17, ATI Industrial Automation, Inc., USA) whilst the force output from the F/T sensor and the voltage output from the forceps tip sensor were acquired. The Nano17 has a resolution of 3.125mN for the calibration adopted in this study which imposes a limit on the minimum force that can be quantified using this calibration method. Three loading and unloading cycles were conducted over a range of 300mN and are shown in Fig. 4. A line of best fit using the least squares method is fitted through this data. Table I provides a summary of the regression ratios and coefficients of determination for the individual loading and unloading curves.

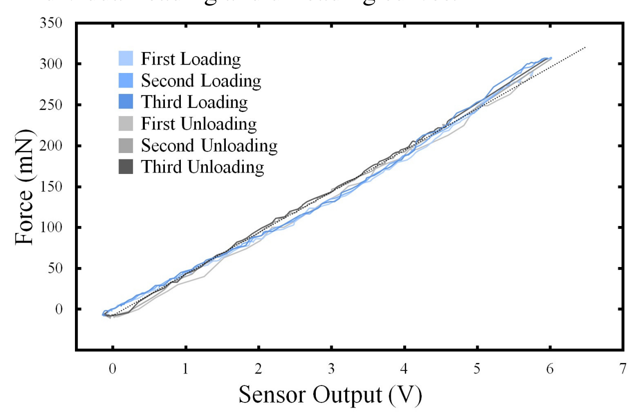

Fig. 4. Calibration experiment showing the voltage output for a force input range of 0-300mN for three successive loading cycles, a linear regression for the entire data set is included on the plot.

TABLE I
TABLE SHOWING THE LINEAR REGRESSION MODELS FOR THE MEASURED GRASPING FORCE SENSOR ASSEMBLY CALIBRATION

	Regression Ratio (mN/V)	Coefficient of Determination R^2
First Loading	50.94	0.999
First Unloading	50.69	0.996
Second Loading	50.77	0.999
Second Unloading	51.29	0.994
Third Loading	50.54	0.999
Third Unloading	50.46	0.993
Mean	50.78	0.997
Standard Deviation	0.30	0.003

The results presented in Fig. 4. and Table I indicate that the sensor assembly produces a repeatable and linear response over a range of 0-300mN. It can be observed that there is a slight hysteresis effect between loading and unloading cycles which is most likely due to the polymer-based material from which the force-sensor assembly is constructed from.

B. *Force-Amplification Bench Tests*

The force-feedback forceps were evaluated in a bench test designed to evaluate the force-amplifying abilities of the system. Measurements of the forceps grasping force F_g and the fingertip force F_f were made as the device was loaded and unloaded for a series of amplification factors between 3-50. The amplification factor is denoted as k and is defined as the ratio F_f / F_g. The calibration described in the previous section was adopted for the F_g measurements. The force-feedback sensor that makes measurements of F_f was calibrated against a precision weight. A spring was used to support the reaction forces generated by the VCA whilst the forceps tip sensor was loaded and unloaded. The results are presented in Fig. 5.

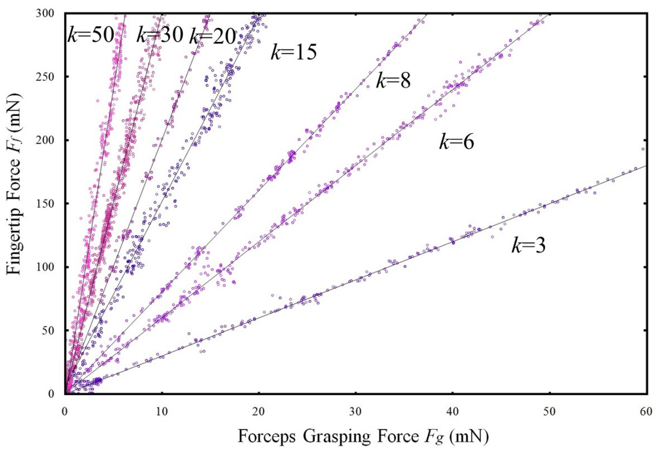

Fig. 5. Plot showing the force-amplifying abilities of the force-feedback forceps for 7 amplification factors through a range of 3-50.

Fig. 5. confirms the force-amplifying abilities of the force-feedback forceps. Lines of best fit have been added to the measured data for the range of amplification factors. These only consider the linear region in which force-amplification occurs, neglecting the discontinuities introduced by the VCA saturating at its peak power consumption. The force-feedback forceps can retain controlled force amplification at all measured amplification factors. Whilst the VCA can generate forces greater than the 300mN range displayed in Fig. 5, this is the most significant region of interest as it represents the perceptible force levels displayed to the fingertip (in the order of $\times 10^2$ mN [12]) against the otherwise imperceptible force levels exerted by the grasping forceps (< $\times 10^1$ mN [8]).

C. *User Study*

A user study was conducted in order to evaluate the effectiveness of the force-feedback forceps in a grasping task with the hypothesis that force-feedback reduces the force threshold at which a user can detect grasping of an object. The experiment compares the force-feedback forceps prototype against standard microsurgical forceps (S&T D-5A, Dumont Surgical, Gaithersburg, USA), the identical model to the forceps that were integrated into the force-feedback prototype. These conventional forceps weigh 15g. Eight users (5 male and 3 female) with an age range of 24 to 33 years were recruited for this study. All users were right-handed and none reported prior clinical experience. The experimental setup is shown in Fig. 6.

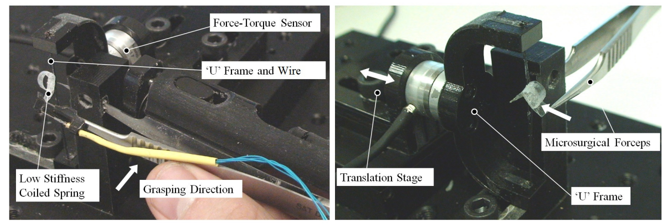

Fig. 6. The experimental setup shown from two angles with the force-feedback forceps prototype (left) and conventional microsurgical forceps (right), the grasping plane is denoted by the white arrows.

The forceps were arranged such that one lever of the forceps was grounded whilst the other lever was free to be grasped by the users. A wire and low stiffness spring were mounted to a 'U' profiled frame so as to be orientated longitudinal to the grasping plane of the forceps and positioned so that contact would be made upon grasping of the forceps by the users. The 'U' profiled frame was coupled to a grounded F/T sensor (Nano17, ATI Industrial Automation, Inc., USA) so that an independent measurement of the forceps grasping force could be made. This arrangement was adopted to ensure the measurement of a bilateral grasping force, otherwise users could potentially detect contact resulting from only a single lever of the forceps. Since it was expected that the force-feedback forceps would be capable of determining considerably lower force thresholds, a low stiffness spring fabricated from polyurethane was placed in series with the wire so that the forceps tip would always make initial contact with the low stiffness spring. A spring of low stiffness was used, firstly to be representative of the soft tissue encountered in microsurgery and secondly, in order to minimize errors as a result of the reaction time between the users making contact and perceiving contact with the spring. Once the soft spring had been fully compressed the forceps would be resisted purely by the wire which was much stiffer, so as to exert up to around 2N at maximum compression of the forceps. This ensured that users would eventually perceive a force in addition to the spring return on the forceps. The tips of the forceps for both cases were mounted so as to contact at the same point along the wire to ensure that it would give the same stiffness response for both experimental set-ups.

The 'U' profiled frame and F/T sensor were mounted on a translation stage that could translate in the direction of the forceps grasping plane. This allowed the position of the wire and low stiffness spring assembly (in the plane of the grasping direction) to be randomly altered to different positions over a range of 3mm offset from the tip of the forceps. This was done in an effort to reduce the user's proprioception sense which could help them learn when they would make grasping contact over the course of a trial. Users were prevented from observing the grasping task so as to rely solely on their sense of touch in detecting the grasping force.

The force data was acquired by the same multi-threaded, real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.) that was used for the control of the instrument. The force data was captured at 50Hz and passed to a host PC. Once a user detected the force, they were requested to immediately stop grasping and depress a push button indicating that they had made contact. This push button generated a binary value which was also logged by the controller and synchronized with the force data. The local maxima of the force measurement, prior to the threshold logging, was taken to be the force detection threshold. The F/T sensor was zeroed before each user trial. The experiment trials were repeated five times for both the conventional and force-feedback forceps. The order in which the trials were conducted was also randomized to prevent biasing of the results. A force-amplification factor of ×30 was used in the trials in which the force-feedback forceps were used. The results of the experiment are presented as a box and whisker plot in Fig. 7. and a summary of the results are provided in Table II.

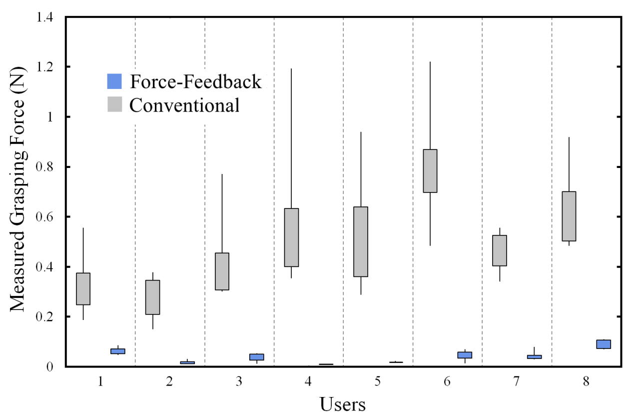

Fig. 7. Box and whisker plot showing the distribution of measured grasping force thresholds for the 8 users in which conventional and force-feedback forceps were used in the experiment.

TABLE II
COMPARISON OF FORCE THRESHOLDS FOR GRASPING WITH CONVENTIONAL AND FORCE-FEEDBACK MICROSURGICAL FORCEPS

	Conventional	Force-Feedback
Mean Force Threshold (mN)	512	41
Maximum Observed Force Threshold (mN)	1221	102
Standard Deviation of Force Thresholds (mN)	256	28
p-value Comparison of Force Thresholds (Kruskal-Wallis)	1.38×10^{-14}	

The results of this study show that the inclusion of amplified force-feedback into microsurgical forceps significantly improves user perception of force, confirming our hypothesis. Both the mean and maximum force thresholds are approximately an order of magnitude higher when conventional forceps are used over the force-feedback prototype. This result can broadly be attributed to the spring return mechanism of the forceps, which hinders perception of the grasping forces. The force required to depress and completely close the tips of the spring-loaded forceps was measured to be 1.7N using the F/T sensor. Even for the grasping displacements that took place during the experiment, it is unsurprising that users could not detect the forces generated by the low stiffness spring that were considerably lower than the baseline forces required to depress the forceps. This is in accordance with the human perceptual response that is described by Weber's law.

All force thresholds measured from experiment using the force-feedback forceps were made before the soft-spring was fully compressed compared to use of the conventional forceps in which users always fully compressed the soft-spring and made contact with the much stiffer wire before perceiving the grasping force. This explains the band separating the highest threshold measured with the force-feedback forceps and the lowest threshold measured with the conventional forceps. The distribution of force thresholds measured with conventional forceps was also greater than with the force-feedback device, with standard deviations of 256mN and 28mN respectively. This too can be attributed to both the 'filtering' effect of the low stiffness spring that ensured users could perceive force before making contact with the higher stiffness wire as well as randomization of the grasping distance which allowed variation in the total amount of force that could be exerted on to the wire.

The use of a higher stiffness wire will have led to errors in the force threshold measurements in the experiment using conventional forceps as a result of the user reaction times. This is because a large force would be generated over a small displacement which would give the users less time to respond to perceiving of the grasping force. Moreover, variations in the speed with which users closed the forceps will have also contributed to the variations in the force thresholds measured. There were also variations in the force threshold perceived between different users which can also be attributed to factors such as variations in the grasping speed as well as the intrinsic perception abilities of different users. However, there is considerably less inter-user variation for the experiment that evaluated the force-feedback forceps, which is also in accordance with Weber's law. It is true that had the users been allowed to complete the study with visual feedback, they would have inevitably been able to perceive grasping contact at much lower force thresholds. However, in bimanual surgical tasks such as micro anastomosis, the surgeon cannot rely on visual feedback throughout if they are to avoid becoming cognitively overburdened.

CONCLUSION

This paper has presented a novel microsurgical forceps design that incorporates amplified force-feedback. The use of a customized force sensor assembly that is sensitive over a range of 0-300mN allows for force amplification factors of up to ×50. This represents a significant improvement upon our previous force-amplifying device [12]. Furthermore, the force-feedback forceps were evaluated in a user study which validated their ability to reduce the force threshold at which grasping can be perceived. Such a device has potential clinical applications in neurosurgery and vascular microsurgery when delicate anatomical structures must be handled by the operating surgeon. Excessive application of force can lead to catastrophic trauma to the patient, for example in the handling of eloquent brain tissue, aneurysms, and during vascular micro anastomosis.

Future derivatives of this device could benefit from many improvements. There is great scope for miniaturization of the design so as to reduce weight and inertia through design optimization and a smaller actuator. This is also important in preventing muscle fatigue in the operator. Future designs that use precision manufacturing techniques would allow miniaturization and facilitate a key step towards producing a clinically-viable device.

The grasping force sensing is crucial in providing effective force-amplification of milli-newton scale force levels and many improvements to the sensor are required in future embodiments. An improved calibration method such as the one adopted in [13] would also provide greater quantification of the force sensing resolution. The addition of a second force sensor to the second forceps tip would also provide redundancy as well as the ability to sense lateral loads in the plane of the grasping. Additionally, sensing of the lateral loads perpendicular to grasping would also prove useful, especially for use in micro anastomosis when the surgeon's perception of lateral loads is important in controlling suture tension. Future work is also required to assess the device's bandwidth and system stability.

In summary, this paper presents force-feedback forceps that have the ability to improve human perception of the delicate grasping tasks encountered during microsurgery.

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