Published in final edited form as:

Int J Med Robot. 2016 March ; 12(1): 85–95. doi:10.1002/rcs.1659.

Hybrid Position/Force Control of an Active Handheld Micromanipulator for Membrane Peeling

Trent S. Wells¹, Sungwook Yang², Robert A. MacLachlan², Louis A. Lobes Jr.³, Joseph N. Martel³, and Cameron N. Riviere^{1,2}

¹Department of Biomedical Engineering, Carnegie Mellon University, Pittsburgh, PA USA 15213

²The Robotics Institute, Carnegie Mellon University, Pittsburgh, PA USA 15213

³Department of Ophthalmology, University of Pittsburgh, Pittsburgh, PA USA 15213

Abstract

Background—Peeling procedures in retinal surgery require micron-scale manipulation and control of sub-tactile forces.

Methods—Hybrid position/force control of an actuated handheld microsurgical instrument is presented as a means for simultaneously improving positioning accuracy and reducing forces to prevent avoidable trauma to tissue. The system response was evaluated, and membrane-peeling trials were performed by four test subjects in both artificial and animal models.

Results—Maximum force was reduced by 56% in both models as compared to position control. No statistically significant effect on procedure duration was observed.

Conclusions—A hybrid position/force control system has been implemented that successfully attenuates forces and minimizes unwanted excursions during microsurgical procedures such as membrane peeling. Results also suggest that improvements in safety using this technique may be attained without increasing the duration of the procedure.

Introduction

Retinal membrane peeling has increasingly been recognized as one of the most important microsurgical maneuvers required to achieve the intraoperative objectives in many retinal surgical diseases such as macular hole, proliferative diabetic retinopathy, and proliferative vitreoretinopathy. Membrane peeling procedures involve peeling epiretinal membranes (ERM), the internal limiting membrane (ILM), the posterior cortical vitreous, or a combination of these from surface of the retina. Unfortunately, membrane peeling often results in some degree of unintended collateral damage to the underlying retina, and since

camr@ri.cmu.edu; +1(412)268-3083.

Disclosures and Ethics

As a requirement of publication author(s) have provided to the publisher signed confirmation of compliance with legal and ethical obligations including but not limited to the following: authorship and contributorship, conflicts of interest, privacy and confidentiality and (where applicable) protection of human and animal research subjects. The authors have read and confirmed their agreement with the ICMJE authorship and conflict of interest criteria. The authors have also confirmed that this article is unique and not under consideration or published in any other publication, and that they have permission from rights holders to reproduce any copyrighted material. Any disclosures are made in this section. The external blind peer reviewers report no conflicts of interest.

the retina does not have a regenerative capacity, such damage may impair its function long-term. Since most visually significant membrane formation and retinal disease occurs in the macula, the region of the retina with photoreceptors most important for fine details in vision, collateral retinal damage from membrane removal is highly significant visually.

The ILM is the most superficial layer of the neurosensory retina, a natural structure that develops with thickness on the order of 3.5 μm . It serves to separate the vitreous body from the retina and guide the nerve fiber layers during development phases (1). An ERM forms in 4–11% of the population over age 50 due to anomalous vitreomacular separation, intraocular inflammation, trauma, or idiopathic causes (2). It consists of a fibrous layer which forms over the ILM ranging in thickness between 33 and 89 μm (3). As the ERM develops, it causes mechanical traction on the retina resulting in distortion in the retinal microstructure and sometimes retinal edema, resulting in impaired function. Rarely, tangential traction of the retina by an ERM may be contributory to development of a macular hole. These conditions impair the patient's vision, sometimes severely to the extent of legal blindness without surgical treatment.

Retinal microsurgeons use tools ranging from 0.5 to 0.9 mm in diameter to access the posterior globe and perform precise manipulations at the level of the retina. Execution of each procedure is influenced by factors such as hand tremor, patient movement, and instrumentation. Failure to properly execute the procedure may lead to complications, the most common in membrane peeling being iatrogenic retinal breaks, intraretinal haemorrhage, and other retinal trauma (4–6). Additionally, failure to achieve the appropriate surgical goals or inadequate membrane peeling may result in recurrence of retinal disease or suboptimal visual improvement. Success depends greatly on a surgeon's previous experience, hand-eye coordination, and fine motor skills.

Recent investigation has identified that peeling the ILM in addition to the ERM can reduce future ERM formation and improve the patient's visual outcome following macular hole repair (7,8). Additionally, for larger macular holes, ILM peeling have become an important component of macular hole surgery and has been shown to improve macular hole closure rates. Surveys taken to identify directions for improvement in ophthalmic procedures have indicated that positioning accuracy and tactile perception have high importance (9). Surgeon accuracy has been reported to be in the range of 50 to 285 μm in various tasks (10–12). This accuracy represents from 0.6 to 8.6 times the thickness of the ERM, and 14 times the ILM thickness at best.

Gupta, Jensen, and de Juan reported the human tactile threshold as 7.5mN (13). However, tearing of the retina is recorded to occur in the range of 7.7 mN in rabbit models (14). Further tests with porcine retina samples have reported tearing between 6.1 mN to 12.4 mN depending on the direction of forces applied (15). In addition, Jagtap and Riviere reported scleral interaction forces tend to be much larger than the forces between the instrument tip and the retina (16). As important as tactile feedback in these procedures may be, these results indicate that tactile feedback without assistance is unachievable for the average surgeon. As a consequence surgeons primarily rely on visual cues and experience to coordinate their movements rather than tactile feedback.

Robotic Assisted Membrane Peeling

Previous approaches to enhancing surgeon positioning and force control have included motion scaling, tremor reduction, haptic feedback, and semi-automated control (17). These approaches have been implemented on multiple robotic systems (18–21); one such platform is the cooperatively-controlled Steady-Hand Eye Robot from Johns Hopkins which uses a stiff, non-backdrivable arm to improve positioning accuracy (22). Force feedback has been developed through auditory cues to the surgeon (22) based on measurements provided by a miniature force sensor (23–25). Several handheld force-controlled devices with one degree of freedom (1DOF) have been developed to provide consistent tissue contact for a variety of applications, including confocal laser endomicroscopy and beating-heart intracardiac surgery (26–29). Multi-degree-of-freedom force compensation has been noted as a future opportunity for handheld medical robots (30); the present paper describes such an approach for minimization of retinal damage during membrane peeling.

Our approach utilizes Micron, a handheld micromanipulator that includes a human-in-the-loop system for increasing positioning accuracy (31). Micron is capable of micron-level detection and positioning as well as force sensing at the millinewton level using a 2DOF force sensor developed at Johns Hopkins University (32). Previous work with Micron focusing on membrane peeling involved vision-based virtual fixtures to limit position and velocity during membrane peeling (33). This approach was successful in limiting applied force; however, the addition of force sensing has potential to yield a more robust system for surgery *in vivo*.

Preliminary results with Micron using hybrid position/force control involved membrane peeling by a single non-surgeon operating on an artificial phantom (34). The present paper describes the addition of a Kalman-based feedforward component to improve force control, and includes experiments with three novices and a trained retinal microsurgeon testing the system in an animal model of retinal peeling *in vivo*, in addition to the previous artificial phantom.

Materials and Methods

Micron, a Fully Handheld Micromanipulator

Micron is a fully handheld 6DOF micromanipulator as shown in Figure 1. The manipulator is a miniature Gough-Stewart platform attached between the end-effector and a cylindrical handle. The platform is actuated by six piezoelectric linear actuators (SQL-RV-1.8 SQUIGGLE® motor, New Scale Technologies, Inc., Victor, N.Y.) operated at 1kHz sampling and capable of reaching a 4mm × 4mm cylindrical workspace (31). The position-control loop in Micron is designed to compensate for normal hand tremor and increase fine positioning accuracy down to the micrometer scale (35). This is achieved by sensing the 6DOF pose of the handle and end-effector, and filtering the measurement to determine the desired motion. The end-effector is then actuated to provide active error compensation, recovering the desired motion. Human hand-eye feedback occurs at approximately 0.5–2 Hz, depending on the task (36,37); our own experience with lowpass filtering in Micron indicates a minimum corner frequency of approximately 1 Hz if destabilization of the hand-

eye feedback loop is to be avoided (35). In this work, a first order lowpass filter at 1.5 Hz is used to reduce tremor.

Position of the manipulator handle and end-effector are sensed at 1 kHz a custom optical tracking system known as ASAP (Apparatus to Sense Accuracy of Position). Three frequency modulated LEDs are mounted on both the handle and end-effector which are detected by two position-sensitive detectors (PSDs) within a 4 cm³ workspace with less than 10µm RMS noise (38).

Millinewton Force Sensor

Direct force feedback is accomplished using a custom 2DOF temperature-compensated force sensor developed at Johns Hopkins University (32). The sensor is constructed from a thin titanium shaft 0.5 mm in diameter, with three optical fibers embedded 120° apart around the circumference of the shaft. Each fiber has a 10-mm long fiber-Bragg grating (FBG) placed 5 mm from its end to measure displacement. A calibration procedure is performed after assembly to correlate the instrument tip displacement to force.

The sensing elements are placed at the distal end of the sensor in order to directly measure the tool-tissue interaction forces. Previous work determined that, in order to accurately sense forces at the retina, the sensor must be placed inside the eye, beyond the incision, in order to avoid the much larger tissue interaction forces at the sclerotomy (16). The overall diameter of the assembly is 0.71 mm to be compatible with standard 25ga ophthalmic surgical instruments. Sensing is accomplished using an optical sensing interrogator (SM130-700 from Micron Optics, Inc., Atlanta, GA) which scans each fiber at 2 kHz to a resolution of 1 pm. This allows a resolution of 0.25 mN to be achieved by the sensor.

Hybrid Position/Force Control

The hybrid position/force control system utilizes Micron's previously developed position-control loop (35) integrated with a direct force-control loop (34). The system is similar to the classic hybrid position/force control originally proposed by Craig and Raibert (39). A notable difference from the classic control method is the lack of the compliance-selection matrix for exclusively defining force or position control. While this can cause instabilities at higher frequencies (40), the slow movements executed during clinical peeling procedures ensure that input stays below roughly 2 Hz (35). The position loop acts to smooth the motion of the end-effector, removing unwanted tremor disturbances. To increase positioning accuracy and regulate applied forces, the position loop is always active, while the force control loop is activated only if the sensed force exceeds a defined threshold. Once active, the necessary position correction from the force loop is added to the filtered output of the position loop. A block diagram of the implemented hybrid position/force control is shown in Figure 2.

Conversion from force sensor coordinates to world coordinates is performed by a rotation matrix based on calibration between the sensor output and the tip movement of Micron. Coordinates are then considered with respect to the handle for purposes of control as shown in Figure 3. A Kalman filter is then implemented to provide a feedforward term for decreased latency and noise within the lower force loop. Gravity compensation in the

control loop is not necessary due to the light weight of the sensor; gravity force is sufficiently mitigated by zeroing the sensor while the manipulator is being held in a peeling orientation.

The force-control loop utilizes three levels of control to provide a seamless integration of the control system during procedures, according to equation (1). A soft and a hard constraint for measured force, F , are specified; for forces below the soft constraint no force control is active. The soft constraint may also be viewed as the force set point. A transition region is specified for forces between the soft constraint, α , and the hard constraint, β ; in this region, force error, $e_f = F - \alpha$, is scaled linearly to provide a corrected force error, e_{fc} . If the sensed force is above the hard constraint, the unmodified force error is sent to the control system in order to mitigate tissue damage.

$$e_{fc} = \begin{cases} 0, & F < \alpha \\ e_f \left(\frac{F - \alpha}{\beta - \alpha} \right), & \alpha \leq F \leq \beta \\ e_f, & F > \beta \end{cases} \quad (1)$$

This corrected force error is then input to the force corrector block (Fig. 2), which is a PID controller that runs at a 1 kHz rate, updating the hybrid position goal H_G according to the gains $K_P=1.4$, $K_I=1.6$, and $K_D=0.3$, with a gain of 0.005 on an exponential decay term for K_I that operates when the force control is not engaged, in order to smooth the transition any time force control is re-engaged.

Artificial Membrane Model

An artificial membrane model was used to provide a repeatable method with which to validate the performance of the control system with a human in the loop. Previous models used to simulate membrane peeling have included New-Skin® Liquid Bandage (41) on various surfaces, Glad® ClingWrap on Sorbothane® rubber (33), and Clear Bandages (23) on PTFE. Sorbothane® rubber was desirable as a substrate for its tissue-like properties (42). New-Skin® Liquid Bandage and Glad® ClingWrap are useful models, but posed difficulties while peeling with a pick due to their stiffness. Clear bandages exhibited a much higher adherence to the rubber surface compared to PTFE, and could not be used.

Polydimethylsiloxane (PDMS) was chosen instead as an artificial retinal membrane model, having been previously used by Gijbels *et al.* in an artificial model for retinal vein cannulation (43). To prepare the model, a 30- μm layer of PDMS was spin-coated onto a silicon wafer. The PDMS was then sectioned into rectangular strips, 5.2 mm $\times$ 20 mm, and transferred onto a layer of Sorbothane® rubber. The width of the strip was chosen to mimic the approximate force exhibited during retinal peeling procedures. A surgeon's feedback about the model confirmed that mechanical behaviour was similar to the ILM. As a check of repeatability, 15 samples were peeled while attempting to maintain a constant velocity, with force measured by the millinewton force sensor described above. A constant velocity was maintained due to the force dependence on rate of peeling. The RMS peeling force was measured to be 5.2 ± 2.2 mN. From these results the soft and hard constraints for peeling were set to 2 mN and 4 mN respectively in both the positive x - and y -directions (direction of

peeling). Along the negative x - and y -directions (opposite the direction of peeling) both constraints were set to zero such that the hard constraint is immediately active.

The setup for the artificial model is depicted in Figure 4. Subjects were trained to peel the PDMS layer from the Sorbothane® rubber while holding the tool orthogonal to the peel direction in order to accurately record force data. In keeping the tool orthogonal to the peeling direction, the y -axis was oriented roughly normal to the surface of the rubber, and the x -axis was aligned roughly with the direction of the peel, as shown in Figure 5.

Animal Model *in vivo*

The animal model setup can be seen in Figure 6. Fertilized chicken eggs were chosen as a membrane-peeling model *in vivo*. This model has been described previously in the literature for simulation of retinal membrane peeling and other retinal manipulations, and is similar to the ERM (44,45). A significant advantage to this model is that damage to the underlying chorioallantoic membrane causes the membrane to bleed, providing an additional realistic biological means for evaluation of the peeling forces.

Newly fertilized eggs were obtained from a local farm (Eichner's Farm Market, Wexford, PA) and incubated until 12 or 13 days old. Trials were performed by removing the eggshell to expose the inner shell membrane (ISM) and the chorioallantoic membrane (CAM). A volume of saline was used to moisten the ISM to simulate conditions within the natural eye. An initial tear was created in the membrane followed by smaller grasping and peeling motions to remove the ISM. As peeling progressed, the egg was rotated to maintain the orientation of Micron within the 4 cm³ workspace of ASAP. The tool was oriented perpendicular to the peeling direction during trials, in order to provide accurate measurement of the forces applied to the tip, since the 2DOF sensor does not sense force parallel to the tool shaft.

Non-surgeon subjects were given significant training (>3 hours) to learn the peeling motions and coordinate the proper orientation of Micron in the ASAP workspace. Each trial consisted of peeling a circular section of the ISM away from the CAM. Force and the number of damaged regions of the CAM (i.e., "retinal tears" in the model) were recorded.

Experimental System

Four subjects performed membrane peeling under a board-approved protocol. One subject was a trained vitreoretinal surgeon with 37 years of surgical experience. The other three subjects were non-clinicians, with varying levels of familiarity with Micron. Membrane-peeling procedures were performed under 16× magnification using a Zeiss OPMI 1 surgical microscope and custom peeling fixtures for the artificial phantom and the egg model (described below). Procedures were also recorded at 30Hz for review through the surgical microscope using two Flea2 1024×768 cameras (Point Grey Research, Inc., Richmond, B.C.).

Since Micron is a fully handheld manipulator, the only stationary coordinates are referenced to ASAP. This does not provide a useful physical relation for evaluating the tasks. To provide relevance for data analysis, all forces are reported with respect to the body frame of

the manipulator, as illustrated in Figure 3. Twelve trials were performed in each model by the four subjects while position and hybrid control modes were randomly varied ($n=24$ for each mode). Welch's t -test was then calculated to determine the statistical significance of the maximum force along each axis, the duration of the trials, and the number of tears observed in the animal model.

Results

The results below are presented for the system response of the hybrid control system under controlled conditions. Further results are presented under more realistic test conditions for both an artificial model and an animal model *in vivo*, comparing hybrid position/force control and with position control only. It should be noted the thresholds have been decreased from previous work in the artificial model (34) to better demonstrate the effectiveness of hybrid control.

System Response

To test the response properties of the hybrid control system without the human in the loop, Micron was clamped in place, and the tip of the manipulator fixed in Sorbothane® rubber, such that any displacement would result in a measured force. The sensor was then zeroed, and a sinusoidal position stimulus was injected at frequencies ranging from 0.5 Hz to 30 Hz. The stimulus or input was generated by commanding a goal vector to the Micron position-control loop, overriding P_G^* in Figure 2. Position noise was eliminated by clamping the handpiece, thus removing the human-in-the-loop component.

Position output (defined as P_M in Figure 2) was recorded by ASAP while injecting the stimulus. With the tip fixed, the force is directly influenced by displacement. The system response was determined by comparing P_M within Figure 2 under the hybrid control to the direct stimulus output. The resulting gain and phase response are shown in Figure 7. The system was determined to have a maximum attenuation of 23 dB, decreasing logarithmically after 0.4 Hz. The system then exhibits a sharper decrease in attenuation after 0.8 Hz. The zero-crossing occurs at 15 Hz and corresponds to a sharp decrease in phase to -180° at frequencies above the crossing. As motion during microsurgical procedures tends to be slow, movement generally occurs at frequencies less than 2 Hz (35), which corresponds to an maximum attenuation of 13 dB. This is sufficient for limiting forces during microsurgical procedures and our peeling trials.

Artificial Membrane Model

The RMS force across all combined trials and the maximum force from each trial were examined under position control and under hybrid position/force control. RMS force was reduced by 32% along the x -axis and 41% along the y -axis. Maximum force across trials under position control was found to be 7.5 ± 2.7 mN along the x -axis and 10.0 ± 3.5 mN along the y -axis. This corresponds to a mean reduction of 46% along the x -axis and 63% along the y -axis, as shown in Figure 8. A significant difference between position and hybrid control was found ($p < 0.05$) along both the x -axis and the y -axis.

The mean duration of trials under position control was 80.7 ± 64.1 s and 69.7 ± 40.4 s under hybrid control; this was not found to be a significant variation ($p = 0.57$).

Histograms of the forces recorded across all trials in the artificial model are presented in Figure 11 for both the x - and y -axis. The occurrences are normalized with respect to the total number of samples recorded.

Animal Model *in vivo*

While training participants in the biological model, soft and hard constraints of 2 mN and 6 mN were chosen along both directions of the x -axis and along the positive y -axis; both constraints were set to zero along the negative y -axis. This deviation from the artificial model is necessary in order to provide the user control over lateral forces as the peel direction is no longer linear for the model *in vivo*. The constraints were found to be sufficient for preventing damage to the underlying CAM. A sample dataset of approximately 28 s duration is given in Figure 9 in which the effect of the hybrid control can be observed. RMS forces were found to be reduced by 34% along the x -axis and 36% along the y -axis. The maximum forces along the x - and y -axis under position control were respectively found to be 15.7 ± 5.5 mN and 12.5 ± 5.3 mN. Under hybrid control the maximum forces were found to be 7.2 ± 1.4 mN and 5.2 ± 1.3 mN along the x - and y -axis, respectively. This corresponds to a significant reduction in the maximum force by 54% along the x -axis ($p < 0.05$) and 58% along the y -axis ($p < 0.05$), as shown in Figure 10. The mean number of tears per procedure of the CAM was found to be 1.4 ± 1.4 without hybrid control and 0.8 ± 1.1 with hybrid control, which did not correspond to a significant improvement ($p = 0.81$). A histogram of the forces recorded across all trials is presented in Figure 12 for both the x - and y -axis. The occurrences are normalized with respect to the total number of samples recorded within the animal model.

No significant difference in the duration of the trials (displayed in Table I) was found in the animal model ($p = 0.72$).

Discussion

The results presented suggest that hybrid position/force control with an active handheld micromanipulator is an effective method for enhancing microsurgical positioning accuracy and control of force. Maximum force was reduced significantly in both peeling models. When computing the vector sum of the axes, total reduction in the RMS force of the artificial model was 36% and 56% in maximum force. A reduction of 31% in total RMS force and 56% in maximum force were observed in the model. This reduction in force may lead to reduced force-related trauma when translated to clinical use. No statistical significance was found in the occurrence of retina tears; however, further investigation is required to determine the cause of the tears. It is speculated that tears were attributable to user inexperience rather than force levels as the experienced users inflicted significantly fewer tears in the animal model as compared to novice users.

In addition, no significant difference was found in the duration of the procedure. Thus a potential increase in patient safety during membrane peeling may be attained without increasing the duration of the procedure.

Future work will include additional benchtop testing with a sclerotomy constraint, followed by testing in animal eyes *in vivo* with greater differentiation between novice and experienced users. In addition, the thresholds will need to be defined for safe operation *in vivo*. Wells et al. (34) examined hybrid position/force control for preventing tears; however, bleeding was observed in chick eggs even in the absence of tears. This suggests that the threshold to avoid trauma should be set lower than the tearing threshold; this requires further consideration.

Matters such as sterilization, electrical isolation, and related safety matters with this instrument remain to be addressed. As a handheld instrument with a rather stiff structure, any failure that required shutting off power would likely still leave Micron functional as a traditional passive instrument.

Acknowledgements

This research was supported by the U.S. National Institutes of Health (grant no. R01EB000526).

References

1. Henrich PB, Monnier CA, Halfter W, Haritoglou C, Strauss RW, Lim RYH, et al. Nanoscale topographic and biomechanical studies of the human internal limiting membrane. *Invest Ophthalmol Vis Sci*. 2012 May; 53(6):2561–2570. [PubMed: 22410559]
2. Johnson, TM.; Johnson, MW. Epiretinal membrane. In: Yanoff, M.; Duker, J., editors. *Ophthalmology*. 3rd ed.. Mosby: St. Louis; 2007. p. 686-690.
3. Wilkins JR, Puliafito CA, Hee MR, Duker JS, Reichel E, Coker JG, et al. Characterization of epiretinal membranes using optical coherence tomography. *Ophthalmology*. 1996; 103(12):2142–2151. [PubMed: 9003350]
4. Jackson TL, Donachie PHJ, Sparrow JM, Johnston RL. Database study of vitreoretinal surgery : report 2, macular hole. *Ophthalmology*. 2013; 120(3):629–634. [PubMed: 23211634]
5. Dogramaci M, Lee EJK, Williamson TH. The incidence and the risk factors for iatrogenic retinal breaks during pars plana vitrectomy. *Eye (Lond)*. 2012 May; 26(5):718–722. [PubMed: 22344186]
6. Nakata K, Ohji M, Ikuno Y, Kusaka S, Gomi F, Tano Y. Sub-retinal hemorrhage during internal limiting membrane peeling for a macular hole. *Graefes Arch Clin Exp Ophthalmol*. 2003 Jul; 241(7):582–584. [PubMed: 12739175]
7. Almony A, Nudleman E, Shah GK, Blinder KJ, Elliott DB, Mittra RA, et al. Techniques, rationale, and outcomes of internal limiting membrane peeling. *Retina*. 2012; 32(5):877–891. [PubMed: 22105502]
8. Haritoglou C, Gass CA, Schaumberger M, Gandorfer A, Ulbig MW. Long-term follow-up after macular hole surgery with internal limiting membrane peeling. *Am J Ophthalmol*. 2002; 134(5): 661–666. [PubMed: 12429240]
9. Huynh N, Akbari M, Loewenstein JI. Tactile feedback in cataract and retinal surgery: a survey-based study. *J Acad Ophthalmol*. 2008; 1(2):79–86.
10. Peral-Gutierrez F, Liao AL, Riviere CN. Static and dynamic accuracy of vitreoretinal surgeons. *Proc Int Conf IEEE Eng Med Biol Soc*. 2004; 4(1):2734–2737.
11. Hotraphinyo, LF.; Riviere, CN. Three-dimensional accuracy assessment of eye surgeons; *Proc Int Conf IEEE Eng Med Biol Soc*; 2001. p. 3458-3461.

12. Wells, TS.; Yang, S.; MacLachlan, RA.; Handa, JT.; Gehlbach, P.; Riviere, C. Comparison of baseline tremor under various microsurgical conditions; Proc IEEE Int Conf Syst Man Cybern; 2013. p. 1482-1487.
13. Gupta PK, Jensen PS, de Juan E Jr. Surgical forces and tactile perception during retinal microsurgery. Lect Notes Comput Sci. 1999; 1679:1218–1225.
14. Sunshine S, Balicki M, He X, Olds K, Kang J. A force-sensing microsurgical instrument that detects forces below human tactile sensation. Retina. 2013; 33(1):200–206. [PubMed: 22810149]
15. Chen K, Rowley AP, Weiland JD. Elastic properties of porcine ocular posterior soft tissues. J Biomed Mater Res A. 2010 May; 93(2):634–645. [PubMed: 19591238]
16. Jagtap, AS.; Riviere, CN. Applied force during vitreoretinal microsurgery with handheld instruments; Proc Int Conf IEEE Eng Med Biol Soc; 2004. p. 2771-2773.
17. Pitcher JD, Wilson JT, Schwartz SD, Hubschman J-P. Robotic eye surgery: past, present, and future. J Comput Sci Syst Biol. 2012; (S3):1–4.
18. Das H, Zak H, Johnson J, Crouch J, Frambach D. Evaluation of a telerobotic system to assist surgeons in microsurgery. Comput Aided Surg. 1999; 4:15–25. [PubMed: 10417827]
19. Ida Y, Sugita N, Ueta T, Tamaki Y, Tanimoto K, Mitsuishi M. Microsurgical robotic system for vitreoretinal surgery. Int J Comput Assist Radiol Surg. 2012; 7(1):27–34. [PubMed: 21573828]
20. Mulgaonkar AP, Hubschman JP, Bourges J-L, Jordan BL, Cham C, Wilson JT, et al. A prototype surgical manipulator for robotic intraocular micro surgery. Stud Health Technol Inform. 2009; 142:215–217. [PubMed: 19377152]
21. Rahimy E, Wilson J, Tsao T-C, Schwartz S, Hubschman J-P. Robot-assisted intraocular surgery: development of the IRISS and feasibility studies in an animal model. Eye (Lond). 2013; 27:972–978. [PubMed: 23722720]
22. Uneri, A.; Balicki, M.; Handa, JT.; Gehlbach, P.; Taylor, R.; Iordachita, I. New Steady-Hand Eye Robot with micro-force sensing for vitreoretinal surgery; Proc IEEE Int Conf Biomed Robot Biomechatron; 2010. p. 814-819.
23. Gonenc, B.; Balicki, MA.; Handa, J.; Gehlbach, P.; Riviere, CN.; Taylor, RH., et al. Preliminary evaluation of a micro-force sensing handheld robot for vitreoretinal surgery; Proc IEEE/RSJ Int Conf Intell Robot Syst; 2012. p. 4125-4130.
24. Balicki M, Uneri A, Iordachita I, Handa J, Gehlbach P, Taylor R. Micro-force sensing in robot assisted membrane peeling for vitreoretinal surgery. Lect Notes Comput Sci. 2010; 6363:303–310.
25. He X, Balicki MA, Kang JU, Gehlbach PL, Handa JT, Taylor RH, et al. Force sensing micro-forceps with integrated fiber bragg grating for vitreoretinal surgery. Proc SPIE 8218, Optical Fibers and Sensors for Medical Diagnostics and Treatment Applications XII. 2012:82180W.
26. Latt WT, Newton RC, Visentini-Scarzanella M, Payne CJ, Noonan DP, Shang J, et al. A hand-held instrument to maintain steady tissue contact during probe-based confocal laser endomicroscopy. IEEE Trans Biomed Eng. 2011 Sep; 58(9):2694–2703. [PubMed: 21768038]
27. Yuen SG, Perrin DP, Vasilyev NV, Nido PJD, Howe RD. Force tracking with feed-forward motion estimation for beating heart surgery. IEEE Trans Robot. 2010; 26(5):888–896.
28. Florez, JM.; Szewczyk, J.; Morel, G. An impedance control strategy for a hand-held instrument to compensate for physiological motion; Proc IEEE Int Conf Rob Autom; 2012. p. 1952-1957.
29. Gilbertson, MW.; Anthony, BW. Ergonomic control strategies for a handheld force-controlled ultrasound probe; Proc IEEE/RSJ Int Conf Intell Robot Syst; 2012. p. 1284-1291.
30. Payne CJ, Yang G-Z. Hand-held medical robots. Ann Biomed Eng. 2014; 42(8):1594–1605. [PubMed: 24927713]
31. Yang S, MacLachlan RA, Riviere CN. Manipulator design and operation for a tremor-manceling microsurgical instrument. IEEE/ASME Trans Mechatron. 2015; 20(2):761–772.
32. Sun Z, Balicki M, Kang J, Handa J, Gehlbach P, Taylor R, et al. A sub-millimetric, 0.25 mN resolution fully integrated fiber-optic force sensing tool for retinal microsurgery. Int J Comput Assist Radiol Surg. 2009 Jun; 4(4):383–390. [PubMed: 20033585]
33. Becker, BC.; MacLachlan, RA.; Lobes, LA.; Riviere, CN. Vision-based retinal membrane peeling with a handheld robot; Proc IEEE Int Conf Robot Autom; 2012. p. 1075-1080.

34. Wells, TS.; MacLachlan, RA.; Riviere, CN. Toward hybrid position/force control for an active handheld micromanipulator; Proc IEEE Int Conf Robot Autom; 2014. p. 772-777.
35. MacLachlan RA, Becker BC, Cuevas Tabares J, Podnar G, Lobes LA Jr, Riviere CN. Micron: an actively stabilized handheld tool for microsurgery. IEEE Trans Robot. 2012; 28(1):195–212. [PubMed: 23028266]
36. Loram ID, Gawthrop PJ, Lakie M. The frequency of human, manual adjustments in balancing an inverted pendulum is constrained by intrinsic physiological factors. J Physiol (Lond). 2006; 577:417–432. [PubMed: 16973712]
37. Kuchen B, Henning K, Rake H, Schafer O. Frequency response of the human controller for various processes and stochastic test signals. Biol Cybern. 1977; 27:33–39. [PubMed: 889956]
38. MacLachlan RA, Riviere CN. High-speed microscale optical tracking using digital frequency-domain multiplexing. IEEE Trans Instrum Meas. 2009; 58(6):1991–2001. [PubMed: 20428484]
39. Craig, JJ.; Raibert, MH. A systematic method of hybrid position/force control of a manipulator; Proc Comput Software and IEEE Comput Soc Third Int Appl Conf; 1979. p. 446-451.
40. Fisher WD, Mujtaba MS. Hybrid position/force control: a correct formulation. Int J Robot Res. 1992; 11(4):299–311.
41. Iyer MN, Han DP. An eye model for practicing vitreoretinal membrane peeling. Arch Ophthalmol. 2006 Jan; 124(1):108–110. [PubMed: 16401792]
42. VanIngen-Dunn, C.; Hurley, TR.; Street, S. Development of a humanlike flesh material for prosthetic limbs; Proc Int Conf IEEE Eng Med Biol Soc; 1993. p. 1313-1314.
43. Gijbels, A.; Wouters, N.; Stalmans, P.; Van Brussel, H.; Reynaerts, D.; Vander Poorten, E. Design and realisation of a novel robotic manipulator for retinal surgery; Proc IEEE/RSJ Int Conf Intell Robot Syst; 2013. p. 3598-3603.
44. Leng T, Miller JM, Bilbao KV, Palanker DV, Huie P, Blumenkranz MS. The chick chorioallantoic membrane as a model tissue for surgical retinal research and simulation. Retina. 2004; 24(3):427–434. [PubMed: 15187666]
45. Kuru, I.; Gonenc, B.; Balicki, M.; Handa, J.; Gehlbach, P.; Taylor, RH., et al. Force sensing micro-forceps for robot assisted retinal surgery; Proc Int Conf IEEE Eng Med Biol Soc; 2012. p. 1401-1404.

Figure 1. Micron, a handheld micromanipulator is shown above with A) millinewton force sensor attached B) Infrared LEDs provide 6 Degree of Freedom tracking of both the handpiece and end effector C) The end effector is actuated by a Gough-Stewart Platform.

Figure 2.

Block diagram of the hybrid position/force control loop implemented on Micron. Hand motion X is the user's position goal P_G corrupted by hand tremor disturbance D_H . This hand motion is mechanically coupled to the output through the manipulator, but the inner position servo loop position loop drives the manipulator output to the position specified by the hybrid goal H_G , using the optically measured position P_M , compensating for manipulator dynamics $G(s)$. This position feedback stabilizes the output in space, rejecting all hand motion. Hand motion only influences the output after passing through a tremor filter, giving an estimate of the user's intended position, P_G^* , which becomes the input to the position servo loop. This stabilization method has been discussed in previous work (26,45). The force control subsystem (presented in this paper) modifies the output position to implement force limit F_L by changing the hybrid goal H_G according to manipulator force F_M detected by the FBG sensor. The force control loop implements a coordinate transform to align the force frame to the position frame and a Kalman filter for feedback filtering. The force filter thresholds the force error according to equation (1); the force corrector is a PID controller that provides a correction in order to provide simultaneous tremor suppression and force control.

Figure 3.

Micron, a handheld micromanipulator and force sensor assembly are shown above. The position coordinate frame is shown in white with axes denoted with 'p' subscript. The force sensor frame is shown in orange at the tip and axes are denoted with subscript 'f'. A coordinate transform is generated after the assembly to align the two frames in the control loop. **Forces presented in this paper are with respect to the body frame (white)** in which the positive y direction will correspond to the primary direction of the peel and the positive x direction corresponds to the left of the tip if holding Micron.

Figure 4.
The surgical setup for artificial membrane peeling experiments showing a) ASAP optical tracker b) surgical microscope c) optical interrogator d) Micron and force sensing pick e) PDMS and Sorbothane[®] layer.

Figure 5.

(top) Peeling of a PDMS layer depicting the force axis in the body frame (bottom) The ideal alignment of the tool and sorbothane layer is depicted. The sensing plane is shown in the body frame with force in the x axis roughly aligned with the direction of the peel and force in the y axis aligned with the surface normal.

Figure 6. Completed peeling procedure shown in the animal model with the Micron tool tip shown in peeling orientation. The top of the egg is removed to expose the ISM and CAM and saline added to moisten the membrane. Peeling the ISM is performed using a surgical microscope. During peeling the forces along the x axis roughly correspond to forces tangential to the CAM while y axis forces are roughly aligned with the surface normal of the CAM.

Figure 7.

System response of the hybrid position/force control loop with the handpiece rigidly clamped. A sinusoidal position displacement was introduced at logarithmically spaced frequencies 0.1 to 20 Hz. A maximum attenuation of 23 dB is observed below 0.4 Hz and decreases to 13 dB at 2 Hz, the maximum expected frequency. The zero crossing occurs at 15 Hz.

Figure 8.

Artificial model peeling results across trials showing a reduction of RMS forces of 32% along the x-axis and 41% along the y-axis (upper) and a reduction in maximum forces by 46% along the x-axis and 63% along the y-axis (lower).

Figure 9.

Sample peeling dataset from the animal model showing force influence under position and hybrid control, the dataset was lowpass filtered at 40 Hz to remove noise. A hard constraint of 6 mN along both the x axis (tangential to CAM) and y axes (normal to CAM) is indicated by the dotted line. The RMS force (bottom) is depicted for both position and hybrid control. Forces under hybrid control can be observed to be much lower in magnitude and stay below the hard constraint of 6 mN, while position controlled peeling frequently exceeds this constraint.

Figure 10.

Peeling results across trials within the animal model showing a reduction of RMS forces of 34% along the *x*-axis and 36% along the *y*-axis (upper) and a reduction in maximum forces by 54% along the *x*-axis and 58% along the *y*-axis (lower).

Figure 11.

Normalized distribution of forces observed along the x -axis (top) and y -axis (bottom) in the artificial model. Higher forces can distinctly be identified as occurring more frequently along both axes under position control. Hybrid control is observed to have predominantly higher frequencies of occurrence in the lower range of forces.

Figure 12.

Normalized distribution of forces observed along the x -axis (top) and y -axis (bottom) in the animal model. Higher forces can be seen to occur more frequently along both axes under position control. Hybrid control can be observed to have predominantly higher frequencies in the lower range of forces.

Table I

Duration of Trials under Position and Hybrid Control for Peeling Trials

Model	Position Control (s)	Hybrid Control (s)
<i>Artificial</i>	80.7 $\pm$ 64.1	69.7 $\pm$ 40.4
<i>Biological</i>	196.6 $\pm$ 100.6	205.4 $\pm$ 70.4