HHS Public Access

Author manuscript

Proc IEEE RAS EMBS Int Conf Biomed Robot Biomechatron. Author manuscript; available in PMC 2018 February 12.

Published in final edited form as:

Proc IEEE RAS EMBS Int Conf Biomed Robot Biomechatron. 2016 June ; 2016: 489–494. doi:10.1109/BIOROB.2016.7523674.

3-DOF Force-Sensing Micro-Forceps for Robot-Assisted Membrane Peeling: Intrinsic Actuation Force Modeling*

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Abstract

Membrane peeling is a challenging procedure in retinal microsurgery, requiring careful manipulation of delicate tissues by using a micro-forceps and exerting very fine forces that are mostly imperceptible to the surgeon. Previously, we developed a micro-forceps with three integrated fiber Bragg grating (FBG) sensors to sense the lateral forces at the instrument's tip. However, importantly this architecture was insufficient to sense the tissue pulling forces along the forceps axis, which may be significant during membrane peeling. Our previous 3-DOF force sensing solutions developed for pick tools are not appropriate for forceps tools due to the motion and intrinsic forces that develop while opening/closing the forceps jaws. This paper presents a new design that adds another FBG attached to the forceps jaws to measure the axial loads. This involves not only the external tool-to-tissue interactions that we need to measure, but also the adverse effect of intrinsic actuation forces that arise due to the elastic deformation of jaws and friction. In this study, through experiments and finite element analyses, we model the intrinsic actuation force. We investigate the effect of the coefficient of friction and material type (stainless

*Research supported in part by the US National Institutes of Health under 2 R01 EB 000526 and in part by Johns Hopkins University internal funds. A. Gao and J. Guo were supported by China Scholarship Council under Grant 201408210100 and 201506020044. A. Gao was also supported in part by National Natural Science Foundation of China under Grant 61473281.

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steel, titanium, nitinol) on this model. Then, the obtained model is used to separate the axial tool-to-tissue forces from the raw sensor measurements. Preliminary experiments and simulation results indicate that the developed linear model based on the actuation displacement is feasible to accurately predict the axial forces at the tool tip.

I. Introduction

Vision-threatening diabetic retinopathy afflicts 8.2% of diabetic adults over 40 in the US [1], or 0.6% of the overall population within this age range [2]. By 2050 the number of Americans with vision-threatening diabetic retinopathy within this age range is estimated to triple [3]. Membrane peeling is an essential and challenging component to the surgical treatment of advanced diabetic retinopathy [4]. It also plays an important role in treating macular puckers/cellophane maculopathy/epiretinal membranes [5], which affects 3.8% of the population aged 45–84 [6], as well as macular holes for which epiretinal membrane and internal limiting membrane peeling may be necessary [7], affects approximately 0.3% of Americans over 50 and can affect both eyes in roughly 10% of patients [8]. Iatrogenic retinal breaks, grasp site injury and hemorrhage due to transmission of injurious forces to the retina are the main risks involved in the procedure [9–12]. Prior work has found that iatrogenic retinal breaks, not related to the sclerotomy, occur in as many as 9.6%–10.7% of cases, and may result in retinal detachments in 1.7%–1.8% of cases [13, 14]. Vitreous hemorrhage can also occur [15], as well as subretinal hemorrhage, which is potentially more serious [16]. Injurious forces transmitted to the retina or supporting tissues that do not manifest as visually observable injury are not yet quantifiable or even identifiable in the usual clinical practice setting. A now commonly performed procedure ILM peeling, has perhaps the steepest learning curve of any intraocular maneuver [17], largely due to the potential for complications resulting from excessive applied forces or traction. The problem is exacerbated by the fact that in the majority of instrument-to-tissue contact events in retinal microsurgery, the forces involved are below the tactile perception threshold of the surgeon [18].

In vitreoretinal microsurgery, during membrane peeling, surgeons routinely use micro-forceps to grasp and peel a very thin layer of fibrous tissue off the retina surface. During the operation, the exerted forces need to be kept at very low levels to avoid retinal injuries. Since most of these forces are imperceptible to surgeons [18], the forces need to be continuously monitored and provided to the surgeon via a proper feedback mechanism during the procedure. In order to measure these very fine forces, sensing instruments with mN accuracy are required. The size and biocompatibility requirements limit the available sensors for this application. The force sensor has to be able to pass through a small incision on the sclera—max. 0.9 mm for 20 Gauge (Ga) tools—so that it can be located close to the tool tip inside of the eye. Only in this way can force measurements pertain solely to the instrument's tip, with no adverse contribution from the sclera [19]. Fiber Bragg grating (FBG) sensors, which are robust optical fibers that detect changes in strain, are able to address these limitations.

Using FBGs, we have previously developed several force-sensing retinal microsurgery tools. First, a single degree of freedom (DOF) force-sensing tool [19] and then a 2-DOF pick-like instrument [20] were built. Intuitively, compared with a pick tool, forceps provide more controlled manipulation of the tissue by firmly grasping it. This enables easier removal of the membrane from the eye in a single step. With this motivation, tool development continued with manual [21] and robot-assisted [22–24] 2-DOF force-sensing forceps tools, which were all reported to provide a reasonably accurate measure of transverse forces. However, the axial force information was not available in any of these designs, which unreasonably required the operator to hold the tool perpendicular to the membrane surface and to peel the membrane by moving only laterally. Considering the sclerotomy constraint and the position of the membrane inside the eye, such motion may not always be feasible. After the membrane edge is grasped, the surgeon predictably applies a Z-axis force (as well as X-Y axis forces) to the membrane to delaminate it off the retina. Therefore, a 2-DOF only force-sensing tool would not accurately represent the relevant forces used. To address this problem, we built a 3-DOF force-sensing pick tool by integrating a central FBG that was shown to successfully capture axial forces within a resolution of 1 mN [25]. However, this solution is not directly transferable to a forceps tool due to the required relative motion between the tool shaft and forceps jaws required to open/close the forceps. Previously, in order to minimize the motion and forces in forceps actuation, we proposed using a customized compliant jaw profile [26], but the complexity in manufacturing and assembly of this tool was beyond our present capabilities.

In this study, we aim at capturing the axial tool-to-tissue interaction forces accurately (within 1 mN resolution) using a simpler and compact embodiment (tool diameter < 0.9 mm), and propose combining our earlier motorized 2-DOF force-sensing forceps [24] with 1-DOF intrinsic force-sensing jaws. In section II, we introduce the design concept of our new tool, and challenges regarding accurate force-sensing. Section III analyses the structure of grasper jaws and proposes two distinct methods to predict the forces generated due to opening/closing action. Section IV shows the experimental results to validate feasibility. Section V investigates the effects of varying friction coefficient and different material properties based on finite element analyses (FEA).

II. Motorized 3-DOF Force-Sensing Micro-Forceps

A. Mechanical Design

For accurate tissue manipulation in retinal microsurgery, involuntary motion, such as the physiological hand tremor, needs to be suppressed. For this, several teleoperated, cooperatively controlled, and handheld robotic systems have already been developed. In order to perform a membrane peeling procedure with any of these systems though requires a proper end-effector that provides firm grasping of the tissue while accurately sensing the tool-to-tissue interaction forces. In our previous work [24], we developed a motorized 2-DOF force-sensing micro-forceps module that can be combined with these systems and works independently from the site of attachment. Fig. 1a illustrates the module mounted on a handheld micromanipulator. The unit is a lightweight "drop-in" module carrying all the

necessary actuators and sensors. The actuation is provided by a linear micro motor (Squiggle-RV-1.8 by New Scale Technologies Inc., Victor, NY), which slides the outer tube down along the tool axis to squeeze and close the compliant forceps jaws as shown in Fig. 1c. The selected micro motor supplies enough force for this motion (0.3 N) in a very small size (2.8×2.8×6 mm), and light weight (0.16 grams).

B. Force Sensing Principle

In order to sense the 2-DOF transverse forces at the tool tip without being influenced by the ambient temperature changes, our previous work [24] employed 3 FBGs fixed evenly around the outer tube (lateral FBGs in Fig. 1a). In this study, we integrate a fourth FBG to detect the forces along the tool axis. This new sensor is located in the center of the outer tube, and is connected to the grasper jaws at the distal end of the instrument. This sensor is located along the neutral axis for—and thus is immune to—side loads at the tool tip. However, it will be affected by (1) the tissue pull/push forces acting along the tool axis and (2) intrinsic actuation forces such as the frictional forces shown in Fig. 1c.

When the outer tube is pushed with a specific force or displacement to close the jaws, it contacts the outer surface of the jaws, developing normal and frictional forces at the contact interface (F_{AI} in Fig. 1c). After the forceps starts interacting with the tissue, such as pulling the membrane to peel it, axial forces are applied at the tool tip (F_{BII}). However, the central FBG is influenced by a combination of all these forces, and the resulting sensor measurement (F_{BI}) will not reflect the correct membrane peeling force. In order to isolate the axial tool-to-tissue interaction force (F_{BII}), a model is required to predict the intrinsic actuation force and subtract it from the raw measurements.

III. Analyses On The Intrinsic Actuation Force

In order to model the intrinsic force during actuation, we followed two methods: an experimental direct force measurement method, and FEA.

A. Experimental Method

We designed two experiments respectively (1) to model the intrinsic actuation force, and (2) to validate the feasibility of this model in separately computing the axial force at the jaw tips. The procedure is summarized in Fig. 2. In this method, two FBGs were calibrated at constant temperature following the procedure in [19] in order to relate the sensors' output to force information. Then in the first experiment, we used medical adhesive Loctite 4013 (tensile strength 28 N/mm²; Henkel, CT) to bond the optical (FBG I in Fig. 3a) and mechanical components (an intermediate tube and forceps jaws). By pulling the jaws incrementally (d_i) and holding the outer tube fixed, we closed the jaws gradually while recording the measured intrinsic actuation force (F_{AI} in Fig. 3a). In the second experiment, while keeping FBG I as before, we attached another force-sensing fiber at the jaws' tips (FBG II in Fig. 3b) and pulled it incrementally (d_j) to introduce gradual external loading (F_{BII} in Fig. 3b). For each external loading condition, forceps jaws were closed/opened incrementally as in the first experiment while recording the output force from both FBG I

and FBG II. In this case, FBG I measurement (F_{BI}) involved the intrinsic actuation force and the external tool tip force (F_{BII}) together. Hypothesizing that the measured force difference between the two FBGs ($F_{BI} - F_{AI}$), and the external tip load are linearly correlated, we solved the optimization problem shown in (1) to find the best fit for coefficient k . The linearity assumption will be validated through experiments in Section IV.

$$\text{minimize} |k \cdot (F_{BI} - F_{AI}) - F_{BII}| \quad (1)$$

B. FEA Method

The goal of this method is twofold: (1) identifying the friction coefficient for the existing geometry and material of the standard forceps jaws, and matching the simulated deformation and actuation forces to the observed experimental behavior in part III.A, (2) based upon the developed finite element model, exploring for ways to reduce the intrinsic actuation force, such as selecting different materials or optimizing the jaw and/or outer tube geometry.

Our design uses components from the standard 23 Ga disposable forceps of Alcon, Inc. (Fort Worth, TX). The outer tube is a 23 Ga (outer tube of 0.57 mm and inner diameter of 0.38 mm) stainless steel tube. The compliant jaws are similarly stainless steel and normally-open. To determine the geometric parameters of the jaws accurately, we captured the projection of jaws, and then used Digimizer (MedCalc Software, Belgium) software to process images. Resulting geometric parameters are shown in Fig. 4.

The identified jaw model with the 23 Ga outer tube was implemented in ABAQUS 6.13 (Dassault Systemes, USA) software to simulate opening/closing action. The material of jaws and outer tube were initially set to be stainless steel, SS316, with the Young's modulus of 193 GPa and Poisson ratio of 0.3. The tube was maintained fixed while an increasing displacement was applied to the proximal end of the jaws to gradually close the jaws. Also, an external axial force was applied to the tip of jaws with varying amplitude.

The contact between the outer tube and jaws as well as the self-contact of jaws occur when the jaws are closed. So for the simulation, three contact conditions were defined. First, the contact between the outer tube and jaws was considered with some friction coefficient. Second, the contact between the tips of jaws was defined with no friction. Third, the self-contact between the inner surfaces of two jaws were considered with no friction. A static analysis was performed for simulating the quasi-static behavior.

IV. Experimental Results

A. FBG Calibration

The experimental setup for FBG calibration is shown in Fig. 5(a). The proximal end of fiber was fixed to the linear ball bearing stage (M-UMR8.25A, Newport, USA), and the distal end was attached with a weight lying on the scale. By lowering the linear stage, the force applied to the FBG was changed from 0g to 50g in 5g increments, while recording the FBG

response. Measurements are shown for FBGs I and II in Fig. 5b, revealing a linear response with a slope of 0.75 and 0.74 respectively.

B. Experimental Modeling and Validation

The setup shown in Fig. 6a was used to conduct two experiments following the steps in Fig. 2. One end of FBG I was attached to the linear ball bearing stage—linear stage I in Fig. 6a—(M-UMR8.25A, Newport, USA), while the other end was fixed to the proximal end of the jaws. The outer tube was held fixed on the optical table. Pulling/pushing the jaws using linear stage I relative to the stationary outer tube causes the jaws to close/open gradually. A nylon thread was glued on the jaw's tips. FBG II was fixed to the nylon thread on one end, while it was clamped onto another stage—linear stage II in Fig. 6a—on the other end. Actuating stage II pulls the nylon thread, and applies increasing force on the jaw tips.

In experiment I, in order to open/close the jaws, we moved linear stage I manually (d_j) from 0 mm (jaws fully open, the outer tube just starting to contact jaws) to 1.8 mm (jaws fully closed) with 0.1mm increments, while the second stage was maintained stationary leaving the nylon thread loose at all times (no force applied on the jaw tips). This motion was repeated 7 times and the measured force along FBG I (F_{AI}) was recorded. As shown in Fig. 6b, the force during actuation rises gradually with an increasing rate. The maximum actuation force is around 225 mN.

In experiment II, linear stage I was actuated manually as in experiment I to open/close jaws. In addition, linear stage II was moved to three different positions (case 1: $d_j=1.5$ mm, case 2: $d_j=1.6$ mm, and case 3: $d_j=1.7$ mm), tensioning the rubber piece in Fig. 6a, and applying varying forces to the distal end of jaws (F_{BI}) via the nylon thread. These forces were detected by FBG II, while the forces due to opening/closing and the external tip loading together affected FBG I readings (F_{BI}). For each position of the nylon thread, the forceps were opened/closed multiple times as in experiment I, while FBG I and II data were recorded. For each experimental condition, 7 trials were performed. Measured forces are shown in Fig. 6c. Using the data (F_{BI} and F_{BII}) of case 3 ($d_j=1.7$ mm) trials—which has the maximum external tip load among all 3 cases—and actuation force data (F_{AI}) from experiment I, Eq. 1 was solved to find the best fit coefficient k as 1.07. Then we evaluated case 1 and case 2 data by computing their mean distances from the $k=1.07$ line (Fig. 6d): 0.14 ± 3.05 mN for case 1 and 0.59 ± 5.36 mN for case 2. Based on the tested force range, these correspond to $0.15\%\pm3.29\%$ and $0.43\%\pm3.93\%$ average deviation for cases 1 and 2 respectively.

V. FEA Results

A. The Effect of Coefficient of Friction

The first aim in FEA method is to tune and identify the friction coefficient between the outer tube and jaws so that the simulation results match the experimentally observed behavior. For this, we ran several simulations modifying the friction coefficient from 0 to 0.16 in increments of 0.02, pulling the jaws through the stationary outer tube gradually, and obtaining the jaw deformation for each pull distance and friction coefficient. The simulation

results on the relationship between the force and displacement at the jaw tips for various friction coefficients are shown in Fig. 7. For bigger values of the friction coefficient, the actuation force is observed to be higher, increasing more rapidly as the jaws are pulled (closed). Among all the simulated cases, the behavior for the friction coefficient of 0.06 (Fig. 7) is the closest to the experimentally obtained data (Fig. 6b). The simulated deformation of jaws for this friction coefficient is shown in Fig. 8.

B. The Effect of Jaw Material

In our micro-forceps tool, a major design goal is to keep the intrinsic actuation force minimal. This is essential for two reasons. First, the forceps actuation is provided by a linear micro motor, as shown earlier in Fig. 1a; and the actuation force directly affects the size of the required actuator as well as the weight of this handheld tool, which all need to be minimized for a surgically feasible instrument. Second, in routine use of micro-forceps in retinal microsurgery, the applied forces at the tool tip remain below 20mN. If the actuation force is much bigger than the external force, then the noise in FBG output due to the contact under large forces may decrease the accuracy in predicting the external forces alone.

A possible method for reducing the actuation force is changing the jaw material. We simulated the actuation force for three possible medical materials: SS316, TC4 and NiTi alloy. The Young's moduli were respectively set as 193 GPa, 113 GPa, 50 GPa for these materials. The results in Fig. 9 show larger actuation forces for materials with higher Young's modulus. The forces for NiTi alloy, which is the softest among the simulated materials, are approximately one fourth of those for SS316 and half of those for TC4.

C. Simulation of External Force Prediction: Fixed Jaw Opening with Varying Force at the Jaw Tips

In membrane peeling in retinal microsurgery, the membrane is first grasped, and then pulled away from the retina. Therefore on the micro-forceps side, the jaws are first closed until the distance between the jaw tips equals approximately the membrane thickness; and then pulling forces are exerted on the jaw tips while keeping the jaw opening fixed. We created a simulation case in order to test if the simulated actuation forces from the previous sections be used to compute the axial forces alone during a membrane peeling procedure. In this simulation, the jaws were maintained fully closed ($d_i=1.8$ mm always). Then pulling forces up to 20 mN were applied in increments of 1 mN equally at the jaw's tips. The resulting forces at the base of jaws in the simulations were recorded. The simulation was run for 3 types of jaw material as in Section V.B.

Shown in Fig. 10 is the obtained relationship between $F_{BI} - F_{AI}$ and F_{BII} , which is linear and does not change much depending on the used material. The proportionality constant (k) is 1.04, 1.05 and 1.05 respectively for SS316, TC4 and NiTi. These are all very close to the experimentally obtained value in Section IV.B ($k=1.07$). This indicates that regardless of the material used, $k \cdot (F_{BI} - F_{AI})$ can closely predict the external tool tip forces.

VI. Conclusion

In this paper, we developed a motorized force-sensing micro-forceps concept for retinal microsurgery, that is able to detect not only transverse forces, but also the axial force. In prior prototypes forces were not measured along the forceps axis, however the forces generated while opening/closing the jaws added onto the actual tissue manipulation forces and prevented their accurate measurement. Our present work addressed this problem by experimentally modeling the actuation forces, extending this model via FEA simulations to different friction and material conditions, and eventually using this model to predict sole tool tip force.

The experimentally measured actuation force required for the standard forceps components are within the force range of the selected micro motor already. However, our simulations have revealed that picking different materials can further reduce this force, decreasing the load on the actuator, causing less noise in raw force readings and improving the accuracy in axial force prediction. Our future aims involve optimizing the jaw material as well as geometry. Upon building the instrument, we will also integrate the lateral FBG readings into the prediction of axial forces to enhance accuracy.

Acknowledgments

B Gonenc is supported in part by the J.W. Marriott Foundation.

P. Gehlbach is supported in part by Research to Prevent Blindness, New York, USA, and gifts by the J. Willard and Alice S. Marriott Foundation, the Gale Trust, Mr. Herb Ehlers, Mr. Bill Wilbur.

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Figure 1.

(a) The motorized 3-DOF force-sensing micro-forceps module mounted on a micromanipulator. The outer tube carries 3 lateral FBGs (yellow) to sense transverse forces. There is an axial FBG (orange) in the center detecting axial forces. (b,c) The outer tube is actuated by a linear micro motor to close the forceps jaws and firmly grasp the membrane. This actuation causes frictional and normal forces at the jaw-outer tube interface (F_{AI}). Pulling the membrane induces axial forces on the tool tip (F_{BII}). The axial FBG will be sensing a combination of these forces (F_{BI}).

Figure 2.

Flowchart of the experimental method: Experiment I is designed to obtain the relationship between jaw position (d_i) and the intrinsic actuation force (F_{AI}). In Experiment II, the actuation force (F_{BI}) is measured in the presence of variable external loading at the tool tip (F_{BII}). Obtained relations are used to find the coefficient k , which can then be used to isolate intrinsic actuation force and predict the pure tool tip loading during actual use of the microforceps.

Figure 3.

Forces measured in the experimental method. (a) In experiment I, the forceps jaws were incrementally pulled (d_i) through the fixed outer tube to gradually close them. FBG I at the proximal end of the jaws was used to measure the intrinsic actuation force (F_{AI}). (b) In experiment II, FBG I was maintained to measure the actuation force (F_{BI}) while FBG II was attached at the distal part to incrementally pull jaw tips (d_j) and introduce varying external force (F_{BII}).

Figure 4.
Geometric parameters of the standard 23 Ga micro-forceps jaws.

Figure 5.

(a) Experimental setup for FBG calibration: The fiber was held in a clamp mounted on a linear stage; a weight was hung at its tip; the tension force along the fiber was increased by raising the weight off the scale gradually. (b) Calibration experiments were repeated for 7 times and the linear fit was found for FBGs I and II.

Figure 6.

(a) Setup for experiments I and II. (b) Force on FBG I (F_{AI}) in experiment I. (c) Forces on FBGs I and II (F_{BI} and F_{BII} respectively) in experiment II. (d) The relationship between $F_{BI} - F_{AI}$ and F_{BII} .

Figure 7.

Comparison of simulation results (blue) for various friction coefficients (from 0 to 0.16 in increments of 0.02) and experimental data (red) for the force on FBG I (F_{AI}), i.e. the actuation force.

Figure 8.

The simulation results showing the jaw deformation and stresses for a friction coefficient of 0.06. Full closure requires a displacement of 1.8 mm relative to the outer tube

Figure 9.

The variation of intrinsic actuation force for three different materials: SS316, TC4 and NiTi.

Figure 10.

The relationship between $F_{BI} - F_{AI}$ and F_{BII} is not affected from the material type ($k=1.04$, 1.05 and 1.05 respectively for SS316, TC4 and NiTi). Computing $k \cdot (F_{BI} - F_{AI})$ closely predicts the external loading at the tool tip in all cases.