

# Haptic Noise Cancellation: Restoring Force Perception in Robotically-Assisted Beating Heart Surgery

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## ABSTRACT

Beating heart surgical methods have the potential to remove the need for the heart-lung machine and its attendant side effects, but must contend with the motion of the heart. Recent research in robotically-assisted surgery has produced a handheld, actuated instrument that can track and compensate for heart motion; however, the reaction forces caused by the actuation mechanism make it difficult for the surgeon to feel the heart during the operation, which can lead to unsafe tissue manipulation. This paper investigates an instrument design that negates reaction forces to the user by moving a counterweight out of phase with the moving mass of the actuator. The resulting instrument retains the tracking and motion compensation abilities of the current instrument, but reduces reaction forces felt by the user by over 80%. Subjects used the new instrument in an *in vitro* beating heart surgical contact task and performance was compared to the previously existing instrument. The new instrument provided a 28% increase in user force sensitivity and improved user reaction times by 51%, indicating that the new instrument greatly enhances force perception in beating heart tasks.

**Keywords:** Beating heart surgery, motion compensation, surgical robotics, force perception

## 1 INTRODUCTION

Nearly 700,000 open-heart procedures are performed annually in the United States. These procedures involve stopping the heart and using the heart-lung machine, a pump that circulates and oxygenates the blood. There are a number of serious side effects associated with the use of the heart-lung machine, such as an increased risk of stroke [6] and long-term neurological dysfunction [17], which has spurred interest in procedures performed on the heart while it is still beating – so-called *beating heart* surgery. However, these procedures are difficult to perform because cardiac motions are too fast for humans to track by hand [2, 4].

For some procedures, beating heart surgery could be aided by a small robotic system. One such procedure is mitral valve annuloplasty, where the anatomical structures of interest largely move along a single axis. This permits the use of a robot with a single degree of freedom [13]. Recent research has developed a handheld, robotic tool to assist the surgeon in performing beating heart mitral valve annuloplasty (Figure 1) [12, 13]; the surgeon can then repair the valve despite its rapid motion. The instrument, called the motion compensation instrument (MCI), tracks the fast motion of the heart tissue and allows the surgeon to operate on the beating heart as if it were motionless. *In vivo* tests confirm its ability to successfully compensate for mitral annulus motion [14, 15].

While this instrument enables a new class of beating heart repair procedures, the current device is hampered by a design in which the surgeon using the device cannot easily feel the heart during the

operation because of inertial forces generated by motion of the actuator (Figure 2A). This masks force sensations that reveal the state of contact between the instrument and the cardiac structures, as well as tissue properties at the instrument tip. This makes it difficult for the surgeon to manipulate moving heart tissue in a safe manner.

One method for rectifying the deficit in haptic perception of the current MCI device would be the use of a teleoperated surgical robotic system. In this scenario, the surgeon would interact with a master controller that would relay motion commands to a slave robot that compensates for the heart tissue motion. A force sensor in the instrument would provide a signal for feedback to the surgeon. While this teleoperated approach has been clinically successful in a number of surgical procedures [3, 1], current systems do not have the necessary instrument speed or force feedback capabilities, and systems with the requisite capabilities would be expensive to develop and use.

An alternative is the development of methods for cancelling within the instrument itself the inertial reaction forces that mask the desired haptic perception. This approach is analogous to noise cancellation in audio systems [5]. This approach presents a different set of challenges than conventional haptic interface design in that the goal is to accurately sense and reproduce not the intended haptic stimulus but rather an interfering haptic signal – in this case, the inertial reaction forces from the motion compensation actuator. Previous work toward tremor compensation for microsurgery also attempts to cancel unwanted motions, but in this case the motions are due to the surgeon’s hands and the goal is improved position accuracy, not enhanced haptic perception [9, 8].

In this work, we investigate the challenges of this haptic noise cancellation approach through the development of a new device called the motion compensation instrument II (MCII). This instrument incorporates a counterweight to cancel inertial forces so that they are not transmitted to the user. With this new instrument, the surgeon is able to operate on the beating heart while retaining force perception; that is, the surgeon is able to perform beating heart surgery with nearly the same force information that would be present if the heart and instrument were stationary. In the following, we first describe the design and characteristics of the MCII. Two

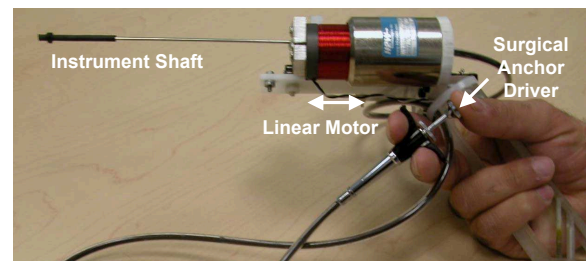


Figure 1: The original motion compensation instrument (MCI) tracks heart structures that move along one axis to assist the surgeon in beating heart surgery [12]. Inertial forces caused by actuation of the motor obscure force perception to the surgeon.

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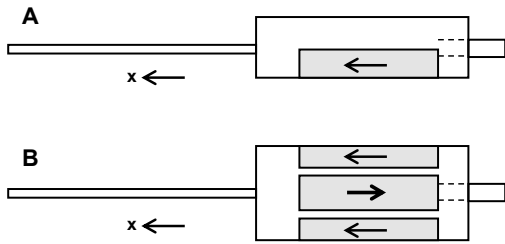


Figure 2: Top view of the MCI actuation mechanism (A) and a counterbalanced actuation mechanism (B). Actuation of the MCI generates inertial forces and torques to the user. In the counterbalanced design (B), reaction forces from the actuator and other moving components (top and bottom arrows) are cancelled by a counterweight moving in the opposite direction (middle arrow). The masses and locations of the moving components are chosen to ensure torque-free actuation.

subsequent user studies comparing the previous MCI with the new MCII in an *in vitro* beating heart task demonstrate that the counterweight design increases force sensitivity and reduces response time for the user.

## 2 COUNTERBALANCED MOTION COMPENSATION INSTRUMENT

### 2.1 Design and Motion Tracking Performance

Like its predecessor, the MCII is intended to compensate for the primarily uniaxial motion of the mitral valve annulus. A successful design must be able to physically track this motion. Adult human mitral annulus motion has been characterized with a maximum velocity of  $210 \text{ mm s}^{-1}$ , maximum acceleration of  $3.8 \text{ m s}^{-2}$ , 18 mm range of motion, and significant spectral components up to at least 10 Hz [13]. The MCII should exceed these specifications using an actuation scheme in which inertial forces are cancelled and no torques about the handle are generated.

These requirements lead to the counterbalanced linear motor design shown schematically in Figure 2B. The actuator mass and other moving components are split into two halves on either side of a counterweight. The counterweight moves 180 degrees out of phase with the actuator to cancel its inertia. The masses and locations of the split components are selected so that torques are not generated when the system moves. The use of a linear motor enables high speed actuation with relatively low moving mass and friction.

The overall design of the MCII is depicted in Figure 3. Actuation of the counterweight is achieved with a capstan that connects the motor slide to the counterweight slide (Figure 4). A capstan is chosen to avoid backlash. The MCII uses a voice coil motor (NCC10-15-023-1X, H2W Technologies, Inc., Valencia, CA, USA) and a high linearity potentiometer (CLP13-15, P3 America, San Diego, CA, USA) for position sensing. These components are mounted on a linear ball-bearing stage (BX4-3, Tusk Direct, Inc., Bethel, CT, USA). The MCII prototype has a 2.54 cm range of motion and is powered by BOP36-1.5 M linear power amplifier (Kepco, Flushing, NY, USA). PID servo control is implemented in a 1 kHz servo loop on a personal computer under Windows XP.

The resulting system has the characteristics required to track the mitral valve annulus. The MCII can attain velocities and accelerations up to  $1.4 \text{ m s}^{-1}$  and  $18.5 \text{ m s}^{-2}$ , respectively. Controller gains were tuned to achieve good stiffness and response. The system is overdamped to avoid dangerous overshoot and instability. The system has a -3 dB point of 18 Hz and roll off rate of 40 dB per decade, which is sufficient to track the mitral valve annulus. The tracking abilities of the MCII were demonstrated by commanding the system to follow human mitral valve annulus motion at 60 beats per

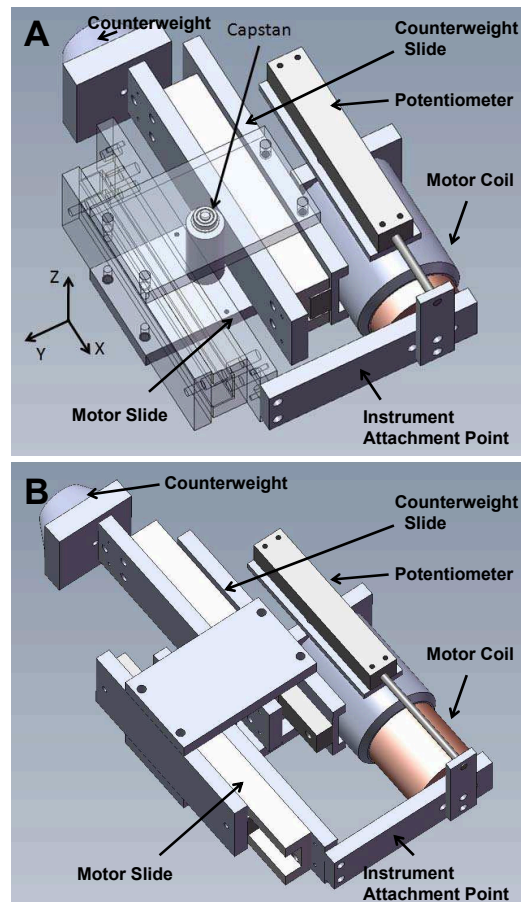


Figure 3: 3D model of MCII retracted (A) and at full extension (B). A handle is mounted to the base (below the middle of the counterweight slide).

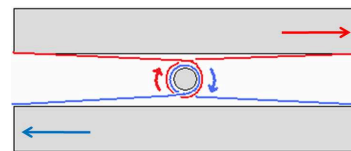


Figure 4: A capstan with two cables. When the top slide moves to the right, the top cable pulls the capstan clockwise. This, in turn, pulls the lower cable toward the left and moves the lower slide to the left. In the MCII, one cable joins the capstan to the motor slide and the other ties the counterweight slide to the capstan.

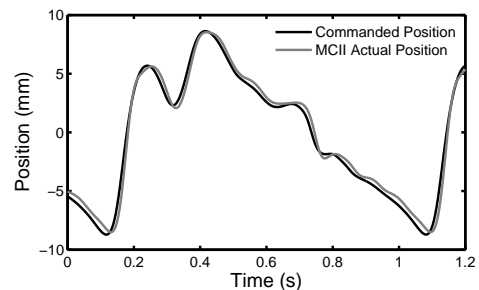


Figure 5: The MCII tracking a prerecorded mitral valve annulus trajectory. Trajectory obtained from [13].

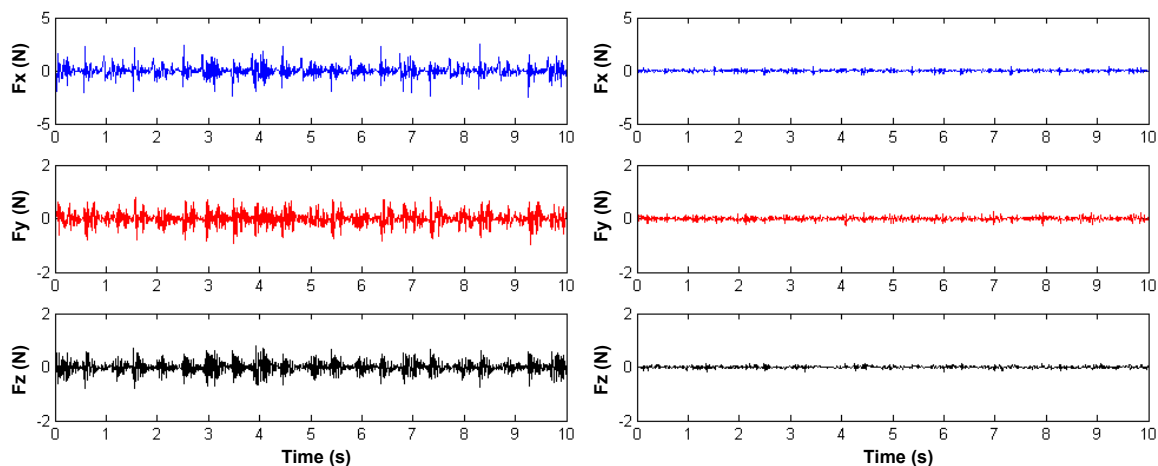


Figure 7: Example force measurements for the MCI (left column) and MCII (right column). Large reaction forces from the moving mass of the MCI are present on all axes. Counterbalancing in the MCII significantly reduces reaction forces.  $F_x$ , the x-axis force, is in line with the linear actuator for both devices.

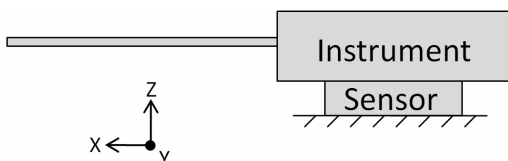


Figure 6: Force-torque characterization setup. The y-axis of the sensor points outward from the page. Both the MCI and MCII were tested in this manner.

minute (bpm) (Figure 5).

Slip is an important consideration in this design. Should the cables slip around the capstan, the backlash-free advantage of this design would be lost. The governing equation for how much tension is required to make a cable slip around a capstan is

$$T_{\text{load}} = T_{\text{hold}} \exp(\mu\theta), \quad (1)$$

where  $T_{\text{load}}$  is the maximum tension that can be sustained on the other side of the capstan before the cable slips,  $T_{\text{hold}}$  is the tension in the cable on one side of the capstan,  $\mu$  is the coefficient of friction between the cable and the capstan, and  $\theta$  is the angle around which the cable is wound. Assuming that the cable is wrapped around the capstan twice ( $\theta = 4\pi$  rad) and both the cable and capstan are made of steel ( $\mu = 0.7$ ), the cable would only slip if the tension on one side of the cable was approximately 6,600 times higher than the other. Thus, with a small amount of tension in the cables, the slip in the system while moving is negligible.

## 2.2 Force-Torque Characterization

A six-axis force-torque analysis was performed to measure the inertial force cancellation properties of the MCII. The handle of the instrument was removed, and a six-axis force-torque sensor (Mini40, ATI Industrial Automation, Apex, NC, USA) was attached to its base, as depicted in Figure 6. The MCII was then controlled to follow a prerecorded trajectory of a human mitral valve annulus, and the resulting forces and torques at the handle were recorded. The MCI was tested similarly to provide a baseline measurement of forces and torques against which to compare the MCII. This was performed three times on both devices for 25 s per trial.

Representative force results for this test are shown in Figure 7, which plots the three axis force data for the MCI and MCII. Figure 8

shows the torques for both devices. The figures indicate that the counterweight in the MCII greatly reduces the forces and torques that result from actuator motion. Figure 9 summarizes the RMS forces and torques across all trials. The RMS forces and torques on most of the axes were reduced by 84% in the MCII. The z-axis torque showed a smaller reduction of 42%. One possible reason for this is that the drive yoke flexes slightly when the motor applies a force on one end. Friction from motion on the slide end of the drive yoke could make the yoke act like a cantilever, and a force on the motor end of this beam could bend it, resulting in a small torque. However, the magnitude of this torque is small (approximately 0.08 Nm peak-to-peak) and may be resolved with a gusset or similar reinforcement between the drive yoke and slide track.

The main observation from this test is that  $F_x$ , the force component along the axis of the instrument and the primary axis of interest to the surgeon during a beating heart operation, was reduced by 84% from 0.58 N (MCI) to 0.09 N (MCII) RMS.

## 3 PERFORMANCE EVALUATION IN A BEATING HEART TASK

Experiments were performed to determine if force perception was actually restored to the user in *in vitro* beating heart tasks. Two studies were conducted in which users were asked to tap a mitral valve annulus motion simulator [13] following a typical annulus trajectory (Figure 5) using both motion compensation devices (Figure 10). The contact surface was a compliant target ( $133 \text{ N m}^{-1}$ ). The first study measured the contact force detection threshold for subjects using the devices. The second study measured the time for subjects to realize and physically respond to making contact with the target when using the devices. In both studies, vision and hearing were obscured so that contact could only be determined by the forces imparted to the user through the hand gripping the instrument handle.

A total of 11 test subjects (ten male and one female, aged 22 to 60; six subjects for the contact force study and five subjects for the contact time study) participated in the studies. Two cardiac surgeons experienced in beating heart surgery and the use of the MCI participated in the contact time study. All subjects participated voluntarily following informed consent under a protocol approved by the University Institutional Review Board.

### 3.1 Contact Force Study

In this study, users were asked to touch the instrument to the simulator while trying to minimize contact forces. Forces were measured

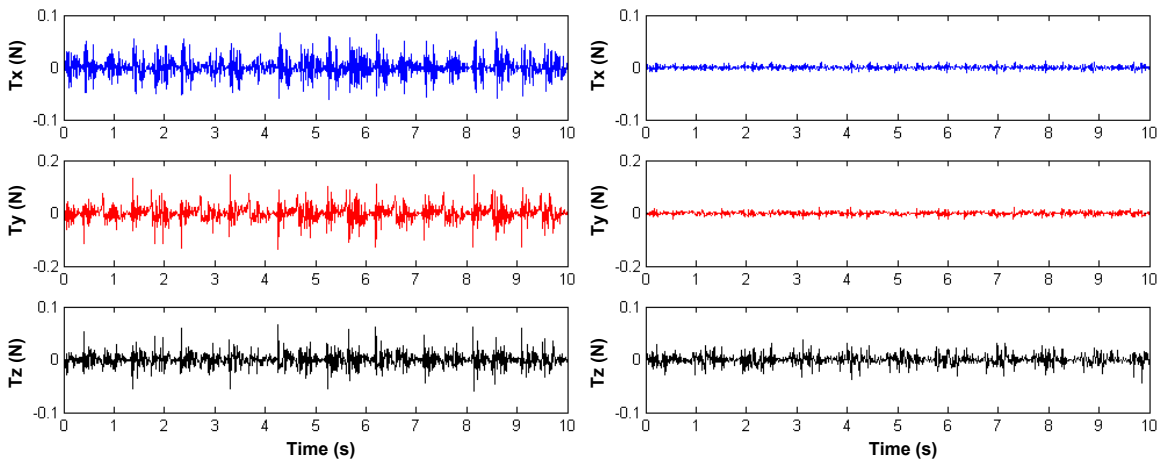


Figure 8: Example torque measurements for the MCI (left column) and MCII (right column). Large torques from the moving mass of the MCI are present on all axes. Counterbalancing in the MCII significantly reduces torques.

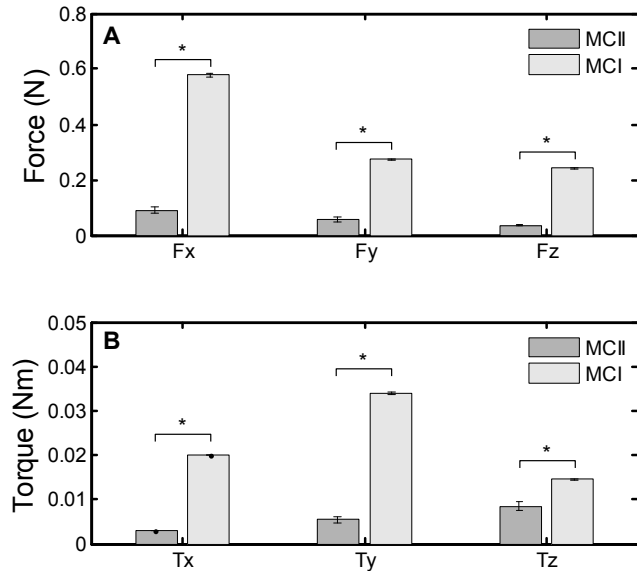


Figure 9: RMS force and torque measurements for the MCII and MCI across three trials. Error bars show standard error. Asterisks indicate statistical significance ( $p < 0.05$ ) between conditions in a two-sided  $t$ -test.

with a custom, tip-mounted optical force sensor (0.17 N RMS accuracy) [16] and recorded.

Users performed the task under six conditions. In the ‘stationary’ condition, the heart motion simulator did not move and the MCII was commanded to a fixed position. This provided a baseline for comparison against trials with a moving target and a motion compensation device. In the ‘MCII’ condition, the heart motion simulator moved at 60 bpm and the MCII tracked its motion. The ‘MCI’ condition did likewise but using the MCI rather than the new MCII device. The remaining three conditions investigated the effect of imperfect instrument balancing by placing incorrect amounts of counterweight on the MCII corresponding to 0.64, 1.59, and 2.12 times the correct mass (208 g).

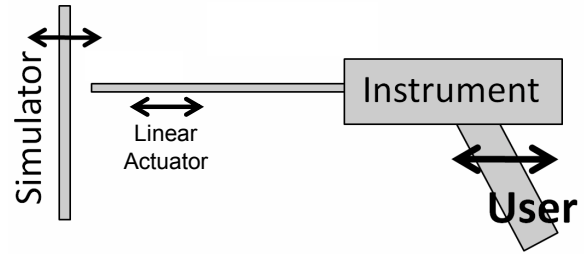


Figure 10: User experiment setup.

### 3.2 Contact Time Study

In this study, users were asked to slowly bring the instrument into contact with the simulator then pull back as soon as contact was felt. Contact was measured electrically using a low voltage circuit that closed when the instrument and simulator touched. Users were asked to perform this tapping task in the same stationary, MCII, and MCI conditions of the contact force study.

The principle behind this experiment was that longer contact times would be indicative of less force perception. For example, contact times should be short when the instrument and the simulator are stationary because users do not have to contend with the motion of the instrument. Contact times with the MCI should be longer than the stationary case because the reaction forces confuse users and make them less able to feel when contact had been achieved. Low contact times with the MCII would indicate that force perception has been restored.

### 3.3 Testing Protocol

Each subject test consisted of a practice period followed by the trials corresponding to their study. Practice familiarized the test subjects with the motion compensation devices and the evaluation task in order to bring the subjects to a uniform level of ability and to limit learning effects during the trials. Practice was divided into three three-minute segments during which the subject was free to experiment with using the MCII to tap the heart motion simulator. During the first segment of training, the target and instrument were stationary. The second segment of training involved a moving target and a stationary instrument. In the third training segment, the target was moving and the MCII tracked its motion.

Following the completion of training, the subjects performed the

trials corresponding to the conditions of their study. In the contact force study, the order of conditions was determined using a balanced Latin square to minimize the effects of between-trial carry-over and learning on collected data. In the contact time study, each user performed the trials in order of the stationary case, the MCI case, and the MCII case. Five trials were performed per condition in both studies, for a total of 30 trials in the contact force study and 15 trials in the contact time study. The means of peak contact forces and contact times were compared for statistically significant differences using Matlab (Version 7.6.0, The MathWorks, Inc., Natick, MA, USA). Comparisons were done by analysis of variance (ANOVA) and planned comparisons between conditions using two-sided  $t$ -tests. In all cases, significance corresponds to  $p < 0.05$ .

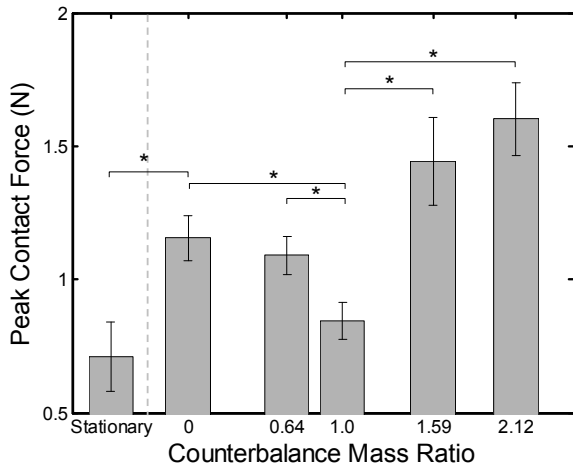


Figure 11: Peak contact forces for the stationary case and over five different counterbalance ratios. A ratio of 0 (no counterweight) corresponds to the MCI. A ratio of 1.0 (perfectly balanced) corresponds to the MCII. All other ratios are the result of mounting the MCII with either less or more weight than required to cancel the actuator inertial forces. Error bars show standard error. Asterisks indicate statistical significance ( $p < 0.05$ ).

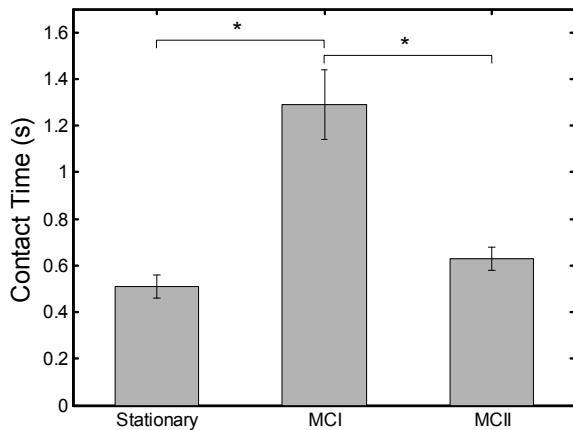


Figure 12: Contact time for the stationary, MCI, and MCII cases. Error bars show standard error. Asterisks indicate statistical significance ( $p < 0.05$ ).

## 4 RESULTS

Figure 11 shows the force sensitivity across users in the contact force study as measured by peak contact forces. The average force for the stationary case was  $0.71 \pm 0.13$  N (mean  $\pm$  standard error). The MCI case (counterbalance ratio of zero) yielded forces that were 63% larger ( $1.16 \pm 0.09$  N) and this difference was statistically significant ( $p < 0.01$ ). In contrast, the corresponding value for the balanced MCII case (ratio of one) was only 18% larger than the stationary case at  $0.84 \pm 0.07$  N. The difference between the MCII and stationary case was not statistically significant for this sample size ( $p = 0.36$ ,  $n = 30$ ). The MCII improved force sensitivity over the MCI by 28% ( $p = 0.007$ ).

An unbalanced instrument (counterbalance ratio not equal to one) led to larger contact forces because the user had to try to detect contact while contending with the inertial forces of the device. Partial balancing (ratio of 0.64) resulted in a 29% contact force increase over the balanced case ( $p = 0.017$ ). Performance was more sensitive to over-weighting the instrument (ratios of 1.59 and 2.12), which increased contact forces by 70–89% ( $p < 0.01$ ).

Figure 12 shows the response time across users in the contact time study. The average contact time for the stationary case was  $0.51 \pm 0.05$  s. The average times for the MCI and MCII cases were  $1.29 \pm 0.15$  s and  $0.63 \pm 0.05$  s (statistically significant difference,  $p = 1.05 \times 10^{-4}$ ). Use of the existing motion compensation device (MCI, central bar) results in reaction times that are 152% longer than the stationary case, with clear statistical significance ( $p = 8.69 \times 10^{-6}$ ). However, with the revised MCII (right-most bar), the mean contact time was only 23% longer. Note that the difference between the MCII and stationary case was not statistically significant for this sample size ( $p = 0.10$ ,  $n = 25$ ). The MCII reduced contact times by 51% when compared to the MCI ( $p = 0.0001$ ). No significant performance differences were observed between the surgeons and nonsurgeons in this study.

## 5 DISCUSSION

In this paper, we investigate the feasibility of active cancellation of “force noise” to enhance haptic perception of contact interactions. We present and validate a counterbalanced motion compensation instrument, the MCII, that restores the force sensation needed to safely manipulate beating heart tissues during surgery. Our results show that the MCII improves user force sensitivity and response time over the existing motion compensation device by 28% and 51%, respectively.

Results from the contact force study also showed that performance has strong dependence on the amount of counterweighting used in the instrument. This is not surprising since the user had to perform the task in the presence of uncancelled inertial forces when the instrument was not balanced. The stronger sensitivity to overweighting that was observed (Figure 11) is probably due to the nonintuitiveness of performing the task when the reaction forces are reversed from the target motion. These findings have implications for future research in inertially-cancelled motion compensation instruments. For example, an alternate design for the MCII could have employed a second motor that works in the opposite direction but receives the same motor currents as the first motor. This approach may not be as effective because any interactions of the instrument with tissue that result in damped motion of the first motor would not be mirrored by the second motor. This would act as an effective over-weighting of the instrument.

This study demonstrated the feasibility of actively canceling haptic interference in a practical application. Generalization to a wider range of applications will bring up a number of issues. Only a single degree of freedom was required for the MCII, and a sensor was not required to measure the motions to be cancelled: the cable transmission directly coupled the counterweight to the instrument motion. The integration of sensing and actuation in multiple degrees



of freedom for motion cancellation may be challenging, although the approach was successfully implemented in handheld devices for microsurgical tremor reduction [9, 8]. In that application, however, the motions and forces were orders of magnitude smaller than for most haptic-based tasks. In general, haptic noise cancellation can draw upon extensive results in machine design, where minimization of vibration is frequently a goal, in part to reduce human exposure to potentially harmful vibrations [10, 7]. The emphasis in the noise cancellation approach, however, must be in defining those aspects of force or motion (frequency, magnitude, direction, etc.) that interfere with haptic perception and sensory-based motor control.

Future work in the development of the MCII will focus on studying the benefits of motion compensation with and without force perception in more complex beating heart surgical tasks such as anchor driving [11] and suturing. Force perception may benefit these tasks by enabling the surgeon to use the mechanical response of the tissue to guide the anchor/needle as it is driven.

## ACKNOWLEDGEMENTS

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