Sensing of physiological hand tremor for real-time compensation in micromanipulation

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SENSING OF PHYSIOLOGICAL HAND TREMOR FOR REAL-TIME COMPENSATION IN MICROMANIPULATION

WIN TUN LATT

School of Mechanical and Aerospace Engineering

A thesis submitted to the Nanyang Technological University
in partial fulfillment of the requirement for the degree of
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List of Symbols and Abbreviations

$A = [a_x \ a_y \ a_z]^T$	Acceleration vector
$A_e = [a_{e_x} \ a_{e_y} \ a_{e_z}]^T$	Vector representing acceleration at the instrument tool tip
$A_i = [a_{ix} \ a_{iy} \ a_{iz}]^T$	Vector representing acceleration at location $\{i\}$
$A_{ij} = [a_{ij_x} \ a_{ij_y} \ a_{ij_z}]^T$	Difference in acceleration readings between accelerometer $\{i\}$ and $\{j\}$
a_k	Instantaneous acceleration at k th sample
a'_k	Attenuated and phase-shifted sensed acceleration at k^{th} sample
$\hat{a}'_k$	Estimate of a_k
BMFLC	Band-limited multiple Fourier linear combiner
DC	Direct current
DOF	Degree-of-freedom
G	Gravity
HPF	High-pass filter
<i>ITrem</i>	A compact hand-held active tremor compensation instrument developed by the author <i>et al.</i>
LPF	Low-pass filter
Micron	A hand-held active tremor compensation instrument developed by Ang and Riviere.
$M_P = [M_{P_x} \ M_{P_y} \ M_{P_z}]^T = -^S P_e$	
$\max(x)$	Maximum value of the operand “x”
p_k	Instantaneous position at k^{th} sample
$\hat{p}_k$	Estimate of p_k
p-p	Peak-to-peak
$P_{ip} = [P_{ip_x} \ P_{ip_y} \ P_{ip_z}]^T$	Displacement vector from location $\{i\}$ to $\{p\}$
PSD	Position sensitive detector

RMS	Root mean square
RMSE	Root mean square error
SNR	Signal-to-noise ratio
STD	Standard deviation
${}^S A_e = \begin{bmatrix} {}^S a_{e_x} & {}^S a_{e_y} & {}^S a_{e_z} \end{bmatrix}^T$	Vector representing acceleration of the tip due to tremor with respect to {S}
${}^S P_e = \begin{bmatrix} {}^S P_{e_x} & {}^S P_{e_y} & {}^S P_{e_z} \end{bmatrix}^T$	Position vector of the instrument tip due to tremor with respect to {S}.
s_k	Desired periodic or quasi-periodic signal which is to be estimated or modeled
{S}	Steady tip frame
t	Time index
T	Sampling period
T_s	Sampling period
WFLC	Weighted Fourier linear combiner
$\alpha = \begin{bmatrix} \alpha_x & \alpha_y & \alpha_z \end{bmatrix}^T$	Angular acceleration vector
$\Omega = \begin{bmatrix} \omega_x & \omega_y & \omega_z \end{bmatrix}^T$	Angular velocity/rate vector
μ	Amplitude weight adaptive gain parameter
μ_o	Frequency weight adaptive gain parameter

Summary

Human hand motion comprises intended and involuntary or erroneous motion, in which physiological tremor is the most common erroneous motion degrading manipulation accuracy in microsurgery. Many engineering approaches exist to improve microsurgery manipulation accuracy and telerobotics is a popular but costly approach. To reduce cost, Micron, a hand-held instrument that can compensate erroneous motion of a surgical tool tip is previously designed and developed.

Micron comprises three main sub-systems: sensing, filtering or estimation, and manipulation. The error compensation performance is reported to suffer from poor quality sensing of the tool motion by the sensing and filtering sub-systems. This dissertation focuses on methods to improve the sensing and estimation quality of tool tip motion due to physiological tremor.

Two hand-held instrument prototypes are designed and built: MicronII and *ITrem*. The manipulation sub-system is neither the author's work nor described in detail. An optical-based micro motion sensing system (M^2S^2), essential in the design, development and evaluation of the hand-held instruments is designed and implemented.

Physiological tremor sensing using only accelerometers is proposed, thereby making the instrument smaller and lighter, and eliminating problems associated with a magnetometer (employed in Micron). Propositions for the location and orientation of accelerometers in the micromanipulation instruments to obtain peak angular sensing resolution are proposed, resulting in a compact sensing sub-system where the sensing resolution, size and weight are significantly improved in *ITrem*.

Angular acceleration errors due to phase and amplitude differences among accelerometer outputs are eliminated by proposing and employing a "lead-lag" compensation method. A method of identifying accelerometer orientation errors is proposed and implemented. Enhancement of real-time adaptive filtering algorithms (WFLC and BMFLC) is proposed to account for phase-lag. Analytical integration is proposed and incorporated with the modified algorithms.

The compensation performance of *ITrem* is evaluated with bench and hand-held tests. In the bench tests, RMS tremulous motion is reduced from 26 μm to 9 μm by about 65% while in the hand-held test, it is reduced by about 40%. Possible causes that prevent the instrument from better tremor compensation, and future work are described.

Chapter 1 : Introduction

1.1. Background

Hand positioning accuracy of humans is limited by involuntary or erroneous motions that are inherent in the hand movement, resulting in errors in the positioning of any object carried out by the hand. The limited positioning accuracy of the hand causes relatively large errors to exist in operations of micromanipulation tasks microsurgical operations and cell manipulation owing to low signal-to-noise ratio (SNR) whereby intended motion, the motion caused by intention and is considered as signal, has about the same magnitude as that of the erroneous motion which is considered as noise. Shakiness or erroneous motion of the hand prevents manipulation of the objects smaller than the swing or peak-to-peak amplitude of the shakiness or the erroneous motion, causing some types of surgical procedures to be infeasible without engineering aids.

Examples of delicate surgical procedures are removal of blood clots that form on the retina and the treatment of retinal vein occlusions [Tang 00; Weiss 98]. The procedure in the treatment of retinal vein occlusions requires a surgeon to inject anticoagulant, a substance or drug that stops blood from clotting, into very small vessels of the retina. The procedure holds the potential of curing a major cause of blindness, estimated to affect 1.6% of persons aged 49 years or older [Mitchell 96]. However, it is generally considered infeasible as it is practically impossible to insert a micropipette into a $\sim 100\text{ }\mu\text{m}$ retinal vein and deliver the anticoagulant without risking tearing the vessel apart. More examples include procedures in neurological surgery and microvascular surgery.

Physiological tremor is believed to contribute substantial erroneous motion of the hand in microsurgery [Wade 82; Harwell 83]. Elble defined it broadly as “*any involuntary, approximately rhythmic, and roughly sinusoidal movement*” [Elble 90]. Physiological tremor results from oscillations due to mechanical (mechanical-reflex) and neuromuscular activity (neurogenic 8 to 12 Hz). Vascular pulsation, room vibration, and transmitted forces are sources of oscillation due to mechanical. Oscillation due to neuromuscular activity is associated with motor unit firing. Frequency of oscillations due to mechanical depends on limb stiffness and inertia and is readily modified by external spring and inertial loads. The oscillation due to neuromuscular activities has a frequency band of 8 to 12 Hz and is independent of limb stiffness and inertia.

It is stated that the largest physiologic oscillations come from mechanical (passive mechanics) [Elble 90]. However, during microsurgery, the oscillations due to mechanical are typically dampened substantially by arm and wrist supports [Patkin 77]. Since oscillations due to mechanical are dampened, the dominant oscillation comes from neuromuscular activities, and hence the frequencies of the oscillations lie in the band of 8-12 Hz.

The instrument tool tip oscillation due to tremor during vitreoretinal microsurgery can be as large as 50 μm peak-to-peak (p-p) in each of the principal axes [Hunter 93]. It has also been reported that the vector magnitude of the instrument tool tip oscillation due to physiological tremor during the removal of epiretinal membrane, the most delicate part of the vitreoretinal microsurgical procedure, is measured to be 38 μm RMS [Singh 02]. The microsurgical tool tip positioning accuracy desired by the surgeons appears to be ± 10 μm [Charles 96].

1.2. Approaches to Improve Manipulation Accuracy in Microsurgery

Various techniques exist for minimization of physiological tremor amplitude during microsurgery. Wrist rests and other supports reduce the tremor. Many surgeons control their sleep, their caffeine intake, alcohol consumption, and avoid strenuous exercises in an attempt to reduce tremor amplitude [Humayun 97; Lakie 94]. Some surgeons take beta blockers, which have been shown to have an attenuating effect [Elman 98]. Apart from the so-called manual techniques, there are several engineering solutions to enhance micromanipulation accuracy in microsurgery, each having its advantages and disadvantages.

1.2.1. Telerobotic systems

The most common approach of engineering solutions to improve micromanipulation accuracy is telerobotic technology [Das 99; Jensen 97; Hunter 93; Schenker 95], where a robotic arm replaces the unstable human hand. In the past, commercially available telerobotic surgery systems included the Zeus® surgical system from Computer Motion, Inc. and the da Vinci® surgical system from Intuitive Surgical,

Inc. Presently, the only commercially available telerobotic surgical system is da Vinci system from Intuitive Surgical, Inc. A picture of the da Vinci telerobotic surgical system is shown in Fig. 1.1. A new telerobotic system named “Advanced Robotic Telemanipulator for Minimally Invasive Surgery (ARTEMIS)” has been being developed in Germany. The feasibility in an experimental endoscopic surgery has been demonstrated, but further technical development is still required for clinical applications [Schurr 00]. Another system called Laprotek is also being developed in Boston, Massachusetts, by endoVia [Marohn 04]. It is reported that it still requires regulatory clearance by FDA.

Fig. 1.1. The da Vinci telerobotic surgical system [Intuitive 08].

Telerobotic approach allows filtering of erroneous motion between master and slave manipulators via motion scaling, which effectively increase the signal-to-noise ratio. This approach, though effective, is costly and obtrusive from the point of view of the surgeon, who is accustomed to treating the patient with his own hands. The da Vinci surgical system costs one million dollars. The expenses of physician training and annual maintenance are about US\$250, 000 and US\$100, 000, respectively [Faust 2007]. An important feature that is lacking in the current telerobotic surgical systems is haptic feedback, the feedback of sensation of touch. Without tactile sensation and force feedback, remote surgeons have to rely solely on visual cues, thereby reducing ease of

use and increasing operator fatigue. Other issues with the telerobotic systems include bulkiness of the system, setup time, and learning time required to use the system.

1.2.2. Steady-hand robotic system

Taylor *et al.* [Taylor 99] have introduced a "steady-hand" robotic approach, in which a robot and a surgeon directly manipulate the same tool, with the robot having high stiffness, and moving along with only those components of the manual input force that are deemed desirable. While this system cannot scale input motion, it has advantages in terms of cost and likelihood of user acceptance. Moreover, it lends the surgeon a "third hand," holding a tool in position while the surgeon performs other tasks with his own two hands.

1.2.3. Hand-held instruments

Transparency, or lack of obtrusiveness, is an important goal in a surgical system, or any human-machine interaction. Despite research into teleoperative surgery, surgeons still prefer direct manual surgery with hand-held instruments, because of the natural and familiar feel it offers the surgeons.

In order to further reduce cost, and to maximize ease of use, user acceptance, and compatibility with current surgical practice, the concept of tremor compensation via a hand-held instrument is proposed by Bose *et al.* [Bose 1992]. However, the proposal does not seem to have been fully implemented. An active tremor compensation hand-held instrument called "Micron" is implemented by Ang and Riviere *et al.* [Ang 01; Ang 04; Riviere 03] trying to keep the instrument size and weight as close as possible to those of existing passive instruments. The instrument consists of three main sub-systems; sensing sub-system, filtering sub-system, and manipulation sub-system. The instrument senses its own motion, estimate the erroneous component of the sensed motion, and manipulate in real-time the surgical tool which is attached to the manipulator at the distal end of the instrument to nullify the erroneous motion of the tool as shown in Fig. 1.2.

Fig. 1.2. Error compensation in microsurgery with Micron, an active handheld instrument that senses its own motion, distinguishes between the desired and erroneous component of the sensed motion, and manipulate its own tip in an equal but opposite motion to cancel the erroneous motion [Ang 04].

Besides the cost and non-obtrusiveness advantage, one important motivation of this approach is its innate 'fail-safe' characteristic. From a safety point of view, the telerobotic and 'steady hand' solutions has potential of doing catastrophic damages should any malfunction happen. The hand-held instrument, on the other hand, with its limited manipulator end effector workspace which is shown in Fig. 1.3 is not much more dangerous than an existing passive microsurgical tool. Furthermore, in the event of extraordinary medical situations or system glitches where surgeon manual intervention is necessary, the bulky robotic systems in the other approaches might have to be removed to get out of the way of the surgery team. There is no such problem in hand-held approach since the surgery is being performed as it usually does; and even if system failure does occur, one may simply switch off the instrument to reduce it to a normal passive instrument.

Fig. 1.3. Manipulator end effector workspace of Micron. (a) End effector workspace together with the instrument, and (b) Close-up view of the workspace. (The manipulator can provide two degree-of-freedom (DOF) rotational motion about X- and Y-axes and 1 DOF translational motion along the tip direction) [Ang 04].

Hand-held instrument's principle in other applications besides microsurgery

Of the engineering solutions described in the previous section, the hand-held approach can be used not only in microsurgery but also in other applications where the performance of manipulation is hampered by the tremor.

Although manipulation sub-system has to be designed specifically based on the application's requirement, sensing sub-system and filtering sub-system can be designed

in general for many applications. Since design of filtering sub-system is mostly done in software, there is no complication for it to be used in other applications. Sensing sub-system can also be used in many applications if it is designed carefully by considering factors such as compactness, unobtrusiveness, self-contained nature, etc of the sensing sub-system.

Weakness of Micron's sensing sub-system

Although Micron has some advantages over other engineering solutions in improving micromanipulation accuracy, it has not reached to its mature stage where it can really be used for tremor compensation mainly due to poor sensing quality. The size of instrument housing required for the sensing sub-system is not small enough for the ease of manipulation. Moreover, the sensing sub-system of Micron is designed only for microsurgery and cell manipulation where environment magnetic field is known. It cannot be used in applications with unknown magnetic environments.

1.3. Objectives

To enable real-time physiological tremor compensation in micromanipulation besides microsurgery, the objective of the research work is set to be as follows.

1) To achieve a non-obtrusive real-time sensing sub-system which has high enough sensing resolution to detect surgical tool motion due to physiological tremor in real-time and is small enough in size and light enough in weight to be fitted in a hand-held instrument for tremor compensation applications besides microsurgery.

2) To achieve high quality estimation of tremulous motion of a surgical tool from the sensed motion of the tool for real-time compensation.

3) To develop a motion sensing system that can track surgical tool motion in micron level. Obviously, this sensing system must have higher sensing resolution and accuracy than that stated in objective 1). However, this system need not necessarily possess real-time and non-obtrusive features. This last objective will have to be achieved first since this sensing system will be used to provide essential assistance in design and development of the sensing sub-system of the tremor compensation instrument as well as to evaluate the sensing sub-system's sensing performance and the instrument's tremor compensation performance

1.4. Scope

The hand-held active tremor compensation instrument *ITrem* comprises three sub-systems: sensing sub-system, filtering or estimation sub-system, and manipulation sub-system. The dissertation focuses on the sensing and filtering sub-systems.

In order to assist in design and development of *ITrem* as well as to evaluate *ITrem*'s performance, an optical-based micro motion sensing system (M^2S^2) is designed and implemented. Although the initial aim of developing M^2S^2 is to provide assistance in design and development of the sensing sub-system, it becomes an essential tool for design and development of the sensing sub-system. Without it, no affirmation of correctness of sensor placement of *ITrem* sensing sub-system could be done.

As required by the objective statement 1), a non-obtrusive real-time sensing sub-system for tremor compensation applications besides microsurgery is designed and developed. In order to enable the use of the sensing sub-system in unknown magnetic environments as well as to achieve non-obtrusive nature, an inertial measurement unit (IMU) consisting only of accelerometers is employed to form the sensing sub-system. Feasibility of sensing tremulous motion of the tool tip using only accelerometer is presented.

Various placements of accelerometers of the IMU are explored to achieve the lowest possible noise while maintaining the instrument size small enough for the ease of manipulation. To differentiate the tremulous motion from the intended motion while maintaining the real-time feature, Weighted-Fourier Linear Combiner (WFLC) or Band-limited Multiple Fourier Linear Combiner (BMFLC), adaptive zero-phase band-pass filters, are employed. Phase-shift and attenuation of the original signal occur at the output of each accelerometer due to the presence of an inherent hardware filter. Introduction to WFLC and BMFLC filters and a method based on these filters to account for the phase-shift and attenuation are presented.

The integration of the sensing sub-system and the manipulation sub-system to form the complete instrument *ITrem* is performed. It should be noted that the manipulator which is employed in the instrument is not the author's work. Evaluation of the tremor compensation performance of *ITrem* is performed and results presented.

1.5. Organization

The remaining chapters of the dissertation are organized as follows.

Chapter 2 describes specification requirements of a sensing system for sensing of hand's physiological tremor, and review of different sensing modalities and a few selected commercially available motion sensing systems.

Chapter 3 describes design, development, and calibration of an optical-based micro motion sensing system (M^2S^2). It then discusses the performance of the system.

Chapter 4 describes physiological tremor sensing using only accelerometers.

Chapter 5 describes propositions for the placement of accelerometers to obtain high angular sensing resolution and designs of placement of accelerometers in sensing sub-system of the instrument based on the propositions described. A study of a hand-held instrument's angular motion due to physiological tremor in representative micromanipulation tasks is also presented.

Chapter 6 describes design of a compact sensing sub-system of the instrument. A method of elimination of phase and magnitude differences among accelerometers to obtain accurate angular motion calculation is also presented.

Chapter 7 presents calibration of accelerometer orientation errors.

Chapter 8 presents real-time estimation and prediction of tremulous motion from attenuated and phase-shifted sensed motion.

In Chapter 9, integration of sensing sub-system with filtering sub-system to obtain tremulous position of the tool tip is described. Manipulation sub-system is briefly presented followed by hardware integration of the sensing sub-system with the manipulation sub-system to form a compact physical instrument, *ITrem*. Finally, preliminary evaluations and results of the tremor compensation performance of the instrument are presented and discussed.

Chapter 10 concludes the dissertation by reviewing the accomplishments and limitations of this research, and suggesting future directions for improvement.

1.6. Outcomes with respect to the Stated Objectives

Objective (1) is achieved by the following endeavors. A compact sensing sub-system in which accelerometers are employed to sense the rigid-body motion of the instrument is designed and developed. Adequate sensing resolution and compactness

result from following the proposed propositions of placement of accelerometers and taking advantage of geometry of surgical tool with respect to the instrument. The sensing sub-system is non-obtrusive since the sensors are of internally-referenced and are light in weight and fixed to the instrument body. Real-time feature is achieved by acquiring the accelerometer outputs through the use of LabVIEW RT real-time operating system running on a PXI computer system. Sensing accuracy of the sub-system is improved by proposing and employing a better identification and compensation method of accelerometer orientation errors. Proposing and employing lead-lag compensation method also reduces angular acceleration errors and hence improves overall sensing accuracy of the sensing sub-system. Overall sensing accuracy is about 85%.

Objective (2) is achieved by proposing modifications to real-time adaptive filtering algorithms (WFLC or BMFLC) which are based on truncated Fourier series and employing the modified algorithms. The modified algorithms can compensate for the phase-lag introduced by the inherent hardware filter at each accelerometer output. Root-mean-square error (RMSE) in real-time estimation of physiological tremor due to the phase-lag is reduced by about 30% with the modified algorithms.

Objective (3) is achieved by designing and developing an optical-based motion sensing system called (M^2S^2) with Position Sensitive Detectors (PSDs) as sensing elements. PSDs are employed since their resolution is determined by signal-to-noise ratio (SNR) of the system while that of CCD or CMOS sensor arrays typically used in commercially available optical-based systems is limited by the number of pixels in the sensor arrays. Adequate resolution of the system (M^2S^2) is achieved by increasing SNR through various measures which are proposed. Adequate accuracy is achieved by calibrating the system with a feedforward neural network. The resolution of the system is 0.9 μm rms while the accuracy of the system is about 99 %.

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Chapter 2 : Motion Sensing Systems

This chapter begins with presentation on specification requirements of physiological tremor sensing for real-time compensation. It then reviews a few different sensing modalities, systems, and sensors both in principle and commercial availability. Based on the review, conclusion on which sensors to be used is drawn in the last section of the chapter.

2.1. Specification Requirements of a Physiological Tremor Sensing System for Real-time Compensation

Characteristics of physiological tremor such as amplitude and frequency band have been well-studied and well-documented in the literature. Physiological tremor amplitude during eye microsurgery is 50 μm peak-to-peak (p-p) in each principal axis [Hunter 93] and its frequency band is from about 8 to 12 Hz [Elble 90]. Although sensing technologies employed in [Hunter 93; Elble 90] to measure the physiological tremor has adequate sensing resolution and bandwidth to capture the tremor, they are not appropriate to be used for real-time compensation via a hand-held instrument.

In order to be able to detect the tremor without attenuating any of its components, the sensing system should have a sensing bandwidth higher than the tremor frequency band. To reduce the tremor amplitude down to about 5 μm via real-time compensation, it is also desirable to have sensing resolution and sensing accuracy to be at least an order smaller than p-p amplitude of the tremor (i.e., $\sim 5 \mu\text{m}$). Also, the system needs to be run at a sampling rate fast enough to detect a change equivalent to the resolution of the system. To determine the sampling rate required to effectively compensate for the tremor, there is a need to know how fast the tremor amplitude changes (i.e., the rate of change of amplitude). Motion equation of tremor based on an assumed sinusoidal tremor profile [Elble 90] is

$$x(t) = X \sin(\omega t) \mu\text{m} \quad (2.1)$$

$$\dot{x}(t) = X\omega \cos(\omega t) \mu\text{m/s} \quad (2.2)$$

$$\ddot{x}(t) = -X\omega^2 \sin(\omega t) \mu\text{m/s}^2 \quad (2.3)$$

From (2.2) and (2.3), maximum velocity and acceleration can be found as:

$$\max(\dot{x}(t)) = X\omega \text{ } \mu\text{m/s} \quad (2.4)$$

$$\max(\ddot{x}(t)) = X\omega^2 \text{ } \mu\text{m/s}^2 \quad (2.5)$$

Putting the numerical values of tremor frequency of $f = 10$ Hz, and tremor amplitude of $X = 25 \text{ } \mu\text{m}$ in (2.4) and (2.5), maximum rate of change of amplitude, $\max(\dot{x}(t))$ and $\max(\ddot{x}(t))$ are found to be as follows.

$$\max(\dot{x}(t)) = 25 \times 2\pi \times 10 = 1.57 \text{ mm/s} \quad (2.6)$$

$$\max(\ddot{x}(t)) = 25 \times (2\pi \times 10)^2 = 98.70 \text{ mm/s}^2 \quad (2.7)$$

To be able to detect the change of $5 \text{ } \mu\text{m}$ for the purpose of real-time compensation, the maximum time allowed, which is the maximum sampling period, is therefore

$$T_{\max} = \frac{5}{\max(\dot{x}(t))} = 3.2 \text{ ms} \quad (2.8)$$

Minimum sampling rate is therefore

$$f_{s \min} = \frac{1}{T_{\max}} = 314 \text{ Hz} \quad (2.9)$$

However, the sensing system which is used to sense the tremor motion for non-compensation purposes such as off-line analysis of tremor motion does not require that high sampling rate because interpolation can be used to obtain motion information at any instant with acceptable accuracy.

Based on the above discussion, the following performance specifications are defined.

- (i) Sensing resolution $> 2.5 \text{ } \mu\text{m}$,
- (ii) Sensing accuracy $> 2.5 \text{ } \mu\text{m}$ (or Sensing error $< 2.5 \text{ } \mu\text{m}$),
- (iii) Bandwidth $> 12 \text{ Hz}$, and
- (iv) Sampling rate $> 400 \text{ Hz}$.

For the sensing system to be deployed in an operating room, its sensing modality must be non-obtrusive to the surgeon and insensitive to interference from other electronic equipments. Other critical implementation considerations would be the size and weight of the sensing system, if it is to be fitted into a handheld device. In the following sections, a review on a few different sensing modalities and systems which are commonly used in motion tracking/sensing applications will be described.

2.2. Externally Referenced Sensing Systems

Sensing modalities of externally referenced sensors include infrared, radio frequency, visible light, ultrasound, electromagnetic wave, etc. Examples of externally referenced motion sensors for indoor applications include the Optotrak, Polaris, and Aurora systems from Northern Digital Inc., the Fastrak and Liberty systems from Polhemus Inc., and the IS-900 system based on the Constellation technology from Intersense, Inc. Most of externally referenced tracking technologies cannot increase the working volume without degrading the signal-to-noise ratio and have interference susceptibility. In this section, a review of a few selected externally referenced motion tracking systems based on different sensing modalities is given.

2.2.1. *Optical tracking systems*

Optical sensors that are typically cameras with CCD, CMOS, or Position Sensitive Detector (PSD) sensors are employed in optical tracking systems. A CCD or CMOS sensor is an array detector receiving an in-focus or out-of-focus image on the focal plane of a camera depending on the application. A PSD is a photodiode that detects the location of the centroid of fallen light on its surface. Any device that estimates centroids is designed to work optimally with out-of-focus imagery. Therefore, PSD works better with out-of-focus imagery than CCD or CMOS cameras.

Two or more features at the same time cannot be determined by a PSD since it can detect only the centroid position of the fallen light on its surface. If, for example, there are two light spots falling on its surface, it cannot determine individual light spot's position on the surface and can only provide centroid position of the two light spots. On the other hand, CCD or CMOS cameras can determine positions of reflected light spots from more than one feature at the same time.

Optical systems typically have good update rates because of the speed of light. The measurement accuracy and resolution tend to decrease with the distance of the mobile object to the sensor (a function of the working volume), because the relative distance between two points on the sensor becomes smaller as the mobile object gets farther away, thus making the points harder to resolve spatially. Optical noise, spurious light, and ambiguity of the sensed surface are sources of errors. Most of the systems use infrared light to minimize some of these effects. Disadvantage of optical tracking

systems is the need to maintain line-of-sight between sensors and features on the mobile object.

Optical systems can be sub-classed into outside-in and inside-out configurations. In the outside-in configuration, the sensors are fastened to the fixed reference. In the inside-out configuration, the sensors are attached to the mobile object whose position and orientation (pose in computer vision terminology) are to be tracked.

Outside-in systems

In outside-in systems, optical sensors are placed on the fixed reference and capture the pose of the mobile object by detecting features that are placed on the object to be tracked.

Optotrack Certus (NDI), Polaris systems (NDI) and VICON systems are examples of outside-in systems. Features which are active markers or passive markers affixed to a tool or object to be tracked are detected by these systems. Active markers are infrared-emitting markers that are activated by an electrical signal and passive markers are spherical, retro-reflective markers that reflect infrared light emitted by illuminators placed on a fixed location. The position and orientation of the tool is calculated based on the information that the position sensor receives from those markers.

Inside-out systems

In an inside-out configuration, sensors are placed on the mobile object and the emitters are on the fixed reference.

The inside-out configuration typically yields higher resolution and accuracy in orientation than the outside-in. The same rotation of a mobile object around a point (e.g. the head of a user rotating around the neck) will produce more displacement on the sensor in an inside-out than in an outside-in configuration (Fig. 2.1).

An inside-out optical tracking system is introduced by Wang *et al.* [Wang 90] in which bearing sensors are placed on the mobile object while the beacons are fixed at locations on the ceiling or walls. HiBall tracking system is an example of commercially available inside-out systems. Specifications of HiBall tracking system are shown in Table 2.1. The HiBall tracking system consists of three main components (Fig. 2.2). A sensing unit called the HiBall is fixed to each user to be tracked. The HiBall unit observes a subsystem of fixed-location infrared LEDs called the Ceiling. The LEDs are in fact entirely located in the Ceiling. Each LED's location on the Ceiling is known.

Communication and synchronization between the host computer and these subsystems is coordinated by the Ceiling-HiBall Interface Board (CIB) [Welch 01].

Fig. 2.1. Rotational resolution comparison between outside-in systems such as shown in (a) and inside-out systems such as in (b); D , difference in position of light spot before and after rotation, is larger in an inside-out system than in an outside-in system.

Fig. 2.2. HiBall Tracking System [Welch 01].

The mathematics that the system uses to compute the position and orientation of the user's head is based on a photogrammetric technique called *space resection by collinearity* [Azuma 91]. Collinearity is a technique which is usually described in introductory photogrammetry textbooks, such as [Wolf 83].

Six PSDs are employed in HiBall Tracking System and hence six primary "camera" views uniformly spaced by 57 degrees. In theory, tracking can be done using any 3 LEDs that one PSD camera can see. In practice, these 3 LEDs should be widely separated from each other, because of the limited resolution of the PSD and other system errors. Tracking using widely-separated LEDs gives more "geometric strength" to the system of equations generated and yields a better solution. Since the lenses have narrow fields of view, the LEDs that one PSD camera unit sees are simply too close together. Some distance is required between the LEDs to get a good solution. If LEDs are separated large enough for high resolution, they cannot be seen by only one PSD camera at the same time. Therefore, more than one PSD camera is required or else a reasonable solution cannot be achieved.

Measurement accuracy and resolution of HiBall system are not acceptable for micromanipulation works although it has wide working area, and relatively high sampling rate and accuracy than other tracking systems.

2.2.2. Magnetic tracking systems

The world's first motion tracking system is a magnetometer or magnetic tracking system which tracks a yaw direction based on the earth's magnetic field. Magnetic tracking systems can be classified into two categories, Alternating Current (AC) magnetic tracking systems and Direct Current (DC) magnetic tracking systems. The former employ alternating current to excite the field generator coil (transmitter) while the latter use direct current.

Principle of operation

In both systems, the magnetic fields are generated by a transmitter consisting of three orthogonal coils of wire wound around a common core. The coils are activated in sequence by the electronics control unit to generate three orthogonal magnetic dipole fields similar in shape to the earth's dipole field but on a much smaller scale. The transmitter is attached to the fixed reference while the receiver is fixed on the mobile object. The receiver composes of three sensors which measure the components of the field's flux received as a consequence of magnetic coupling. Based on these measurements, the position and orientation of the receiver with respect to the transmitter attached to the reference are determined by the system [Raab 77; Raab 79].

For the AC systems, the transmitter is activated with oscillating currents of 7-14 kHz, and the sensor consists of a tri-axial coil assembly that is able to measure the components of these oscillating magnetic fields by inductive pickup [Foxlin 02]. In the presence of metallic objects, AC magnetic tracking systems suffer from the problem of eddy current distortion. The alternating magnetic field generated by the transmitter induces eddy currents in metallic objects in the environment. The eddy currents, in turn, generate their own magnetic fields, which interact with the magnetic field of the transmitter and distort it. As a result of this distortion, inaccurate position and orientation are reported by the systems in the vicinity of metallic objects.

For the DC systems, the triaxial sensor assembly uses devices sensitive to small DC magnetic fields, namely any of the sensor types that are used in electronic compasses. Traditionally the most popular such sensor is the fluxgate magnetometer, but recently solid-state technologies such as magneto-resistive, magneto-inductive, micromechanical and hall effect sensors have been replacing fluxgates. DC tracking systems avoid eddy currents by using a static magnetic field. They supply short DC

pulses to the transmitter. Once the disturbances caused by the rising edge of a DC pulse settle, the magnetic field generated by the transmitter remains static until the end of the pulse. The static field is measured by the receiver.

Nixon *et al.* [Nixon 98] analyze the effects of metals on AC and DC tracking systems, try different types of metals, and find that the DC tracking system is completely unaffected by brass, aluminum and stainless steel, but has larger errors than the AC tracking system in the presence of copper, ferrite and mild steel. Presumably the AC tracking system performs better with ferromagnetic steel because the magnetic permeability decreases with frequency.

Resolution and accuracy of both AC and DC magnetic tracking systems are inversely proportional to the separation distance between the transmitter and the receiver. To get good resolution and accuracy, the transmitter should be close to the receiver and metals far away from both of them.

Advantages and disadvantages

Magnetic spatial measurement systems do not require line-of-sight clearance. Regardless of AC or DC types, all magnetic spatial measurement systems are affected by the close presence of ferromagnetic materials. Ferromagnetic materials distort the generated magnetic fields, leading to incorrect sensor readings.

A series of DC tracking systems has been produced and marketed by Ascension Technology. This includes microBIRD, miniBIRD, pciBIRD, and Flock of Birds. Although the sensor sizes are small enough to be put inside a hand-held instrument, the accuracy and resolution requirements for micromanipulation works cannot be fulfilled apart from susceptibility to magnetic interferences. Other systems that employ electromagnetic modalities include Aurora (Northern Digital Inc), Fastrak (Polhemus, Inc), Patriot (Polhemus, Inc), and Liberty (Polhemus, Inc), etc.

2.2.3. Ultrasonic tracking systems

Principle of operation

Time-of-flight (TOF) technique is employed in commonly used ultrasonic tracking systems and three or more ultrasonic transmitters placed on the mobile object and three or more receivers fixed on the reference are involved in the systems [Logitech

91]. However, transmitters can also be placed on the reference while receivers on the mobile object [Foxlin 98b].

The relative spatial positions of each transmitter on the mobile object and of each receiver on the reference are known. The transmitters emit ultrasonic pulses at regular intervals with only one transmitter emitting at any given time. The receivers detect these pulses and record their times of arrival, which are then used to compute the distances between the transmitters and the receivers. These distances and the known positions of receivers constitute sufficient information to get the pose of the mobile object.

Advantages and disadvantages

There is no magnetic interference problem. The transmitter unit held by the user is small and lightweight. Maintenance of line-of-sight between the transmitter and the receiver is necessary. Accuracy is affected by the constancy of the velocity of sound in air. The low update rate results from the sequential triple emission of sound signals and the low speed of the sound. Echoes and external noise create spurious measurements, leading to inaccuracies in the reported position and orientation.

2.2.4. Summary of commercially available motion tracking systems

The sensing modality, size, weight, and performance specifications of some of the selected commercial externally referenced motion sensing systems are summarized in Table 2.1. All of the systems require their sensing module(s)/device(s) to be fitted on the mobile object. As can be seen from the table, all systems mentioned cannot provide sufficient resolution and accuracy required to detect physiological tremor as specified in section 2.1.

Table 2.1 Summary of selected commercially available motion sensing systems

System	Optotrak Certus (NDI)	Polaris (NDI)	Aurora (NDI)	Fastrak (Polhemus, Inc)	Liberty (Polhemus, Inc)	Patriot (Polhemus, Inc)	Constellation IS-900 (Intersense, Inc)	HiBall-3100 (3rdTech, Inc)	MicroBIRD (Ascension Technology)
Modality	Infrared	Infrared	Electro-magnetic	Electro-magnetic (AC)	Electro-magnetic (AC)	Electro-magnetic	Ultrasound + Inertial	Infrared	Electro-magnetic (DC)
Sampling Rate (Hz)	1500	60	40	120	240	60	180	2000	90
Accuracy (RMS)	deg	NA	0.4	0.15	0.15	0.75	0.25	0.02	0.5
	µm	250	600	760	760	2540	2000	400	1400
Resolution	deg	NA	0.2	0.3 @ 0.3 m	0.00012 @ 0.3m	1.2 deg @ 0.3 m	0.05	0.01	0.1
	µm	NA	400	60 @ 0.3 m	38 @ 0.3m	450 @ 0.3 m	750	200	500
Weight (gram)	<5	<10	0.7	NA	23	NA	44	170	NA
Size (mm)	Ø4 x 2 per unit	65 x 65 x 8	9 x Ø1.8	22.9 x 27.9 x 15.2	22.9 x 27.9 x 15.2	22.9 x 27.9 x 15.2	100 x 80 x 44	73 x Ø 54	6.5 x Ø1.3

2.3. Internally Referenced Sensing Systems

Inertial sensors such as accelerometers and gyroscopes are examples of internally referenced sensors. Internally referenced sensors are self-contained since an external reference is not necessarily required to obtain motion information of a mobile object to which the sensors are attached.

A sensing system consisting of internally referenced sensors can be termed as an internally referenced sensing system. An example of internally referenced sensing system is an inertial measurement unit (IMU) which consists usually of three accelerometers and three gyroscopes.

With well-defined initial conditions, a six degrees-of-freedom (DOF) IMU can measure the motion of a body in three dimensional Euclidean space – acceleration, velocity, position, angular rate, and orientation (or attitude in navigation terminology).

2.3.1. Accelerometers

An accelerometer is a precision instrument designed to measure a component of the vector quantity called specific force, f . It is defined by the equation

$$f = a - g$$

where a is inertial acceleration and g is gravitational attraction per unit mass. In navigation computation, g must be determined and $f + g$ must be used to obtain the inertial acceleration a of a moving object.

Numerous technologies are used to implement today's accelerometer designs, including piezoelectric, piezoresistive, and capacitive technologies. Specifications of representative commercial accelerometer models are shown in Table 2.2.

Table 2.2. Summary of a few selected low-g commercial accelerometers.

Make/Model	Type	No. of Axes	Input Range (g*)	Cost range (\$)	Size (mm)	Sensitivity (mV/g*)	Weight (grams)	Output Noise @ 12 Hz(mg*)	Bandwidth (Hz)
Kistler / 8393A2	Capacitive	3	2	20	31.8 × 31.8 × 8.9	1000	60	0.13	0 to 300 (%5)
Analog Devices/ADXL203	Differential capacitive	2	1.7	12	5 × 5 × 2	1000	1	0.5	0.5 to 2500 #
Analog Devices/ADXL330	Differential capacitive	3	3.6	5 to 10	4 × 4 × 1.45	300	1	1.2	0.5 to 1600 #
MEMSIC/ MXR9500 G/M	heat transfer	3	1.5	NA	7 × 7 × 1.8	500	NA	2.1	17 (3 dB)
Measurement Speacilities(AMP)/ ACH-04-08-05	Piezo-electric	3	250	25-50	10.16 × 10.16 × 1.5	1.8	0.35	0.7	0.35 to 5000 (3 dB)
Entran/EGAX	Piezo-resistive	1	10	500	3.56 × 3.56 × 6.9	8	NA	NA	0 to >250 (0.5 dB)
Entran/EGAS	Piezo-resistive	1	5	NA	5 × 5 × 10	15	<1	NA	0 to >150
IC sensors/3021	Piezo-resistive	1	10	NA	15.24 × 15.24 × 5	6	1.4	0.33	1000(%5)

g* denotes gravity

Depends on the value of output capacitor used.

Regardless of sensing mechanism, the vast majority of modern accelerometers are of the pendulous type [Verplaetse 96]. Pendulous accelerometers feature an inertial proof mass, a segment of the sensor with known mass, that is mechanically coupled to the rest of the sensor by a spring-like hinge or tether—a cantilever structure is an example of a pendulous system. In this type of accelerometer, when the sensor is accelerated from rest, its proof mass tends to stay at rest, and the spring is deformed. It is this deformation of the spring (or the corresponding displacement of the proof mass) that is transduced to become the output signal of the sensor. Capacitive and differential capacitive accelerometers generally have higher sensitivities than piezoresistive models.

The MEMSIC device operation is based on heat transfer by natural convection and operates like other accelerometers having a proof mass except it is a gas in MEMSIC sensor. Heat source, centered in the silicon chip is suspended across a cavity. Equally spaced aluminum/polysilicon thermopiles (groups of thermocouples) are located equidistantly on all four sides of the heat source. Under zero acceleration, a temperature gradient is symmetrical about the heat source, so that the temperature is the same at all four thermopiles, causing them to output the same voltage. Acceleration in any direction will disturb the temperature profile, due to free convection heat transfer, causing it to be asymmetrical. The temperature, and hence voltage output of the four thermopiles will then be different. The differential voltage at the thermopile outputs is directly proportional to the acceleration.

2.3.2. Gyroscopes

Gyroscopes directly measure angular velocity/rate of an object onto which they are mounted, without the need for special calculations. Gyroscopes are also referred to as “gyros” or “rate gyros” or “angular rate sensors”.

In general, gyroscopes can be classified into three different categories based on their performance: inertial-grade, tactical-grade, and rate-grade devices [Yazdi98]. Table below summarizes the requirements for each of these categories [Lefevre93; Lawrence93].

Table 2.3 Performance specifications of different grade gyros [Yazdi 98]

Parameter	Rate grade	Tactical grade	Inertial grade
Angle random walk, $\frac{^\circ}{\sqrt{h}}$	> 0.5	0.5-0.05	< 0.001
Bias drift, $\frac{^\circ}{h}$	10-1000	0.1-10	<0.01
Scale factor accuracy, %	0.1-1	0.01-0.1	<0.001
Full scale range ($\frac{^\circ}{s}$)	50-1000	> 500	> 400
Max. shock in 1 ms, g's	10^3	10^3 - 10^4	10^3
Bandwidth, Hz	> 70	~100	~100

Today, optical gyroscopes are the most accurate gyroscopes available in the market. Among these, ring laser gyroscopes have demonstrated inertial-grade performance, while fiber-optic gyroscopes are mainly used in tactical-grade applications. A ring laser gyroscope uses interference of laser light within a bulk optic ring to detect changes in orientation and spin. Although highly accurate, these devices are too expensive and bulky for many low-cost applications. A photo and a schematic diagram of a ring laser gyroscope are shown in Fig. 2.3 and Fig. 2.4 respectively.

Fig. 2.3. Photograph of a ring laser gyro [King 98].

Fig. 2.4. Schematic diagram of a ring laser gyroscope [Titterton 04]

A modern strapdown ring laser gyroscope inertial navigation unit is about 7 x 7 x 11 inches (178 x 178 x 279 mm) in size, weighs about 10 kg, with a power dissipation of about 50 W. It costs some tens of thousands of pounds [King 98].

Micromachined silicon rate-grade gyroscopes, such as those for automotive applications, cost about US\$10–\$20 with a resolution of 0.1 deg/s in a bandwidth of 50 Hz [Yazdi 98]. They are relatively small in size and light in weight and are suitable for low-cost or size-constraint applications such as a hand-held tremor compensation instrument. Table 2.4 shows summary of a few selected commercial rate gyroscopes [Ang 04].

Table 2.4 Summary of a few selected commercial rate gyroscopes [Ang 04]

Make/Model	Honeywell / GG-1342	KVH / DSP-3000	Tokin / CG-L43D	Analog Devices / ADXRS150
Class	Inertial grade	Tactical/Navigation grade	Rate grade	Rate grade
Type	Ring laser	Interferometric fiber optic	Vibrating piezoelectric	MEMS
Cost range	> \$ 10,000	> \$5,000	\$50-100	\$ 30-40
Output noise (@ 15 Hz)	1.3 deg/h	60 deg/h	1.4 deg/s	0.75 deg/s
Bias stability (deg/h)	0.0035	20	NA	NA
-3 dB Bandwidth (Hz)	NA	50	20	40
Sampling rate	Limited by ADC or microprocessor, usually > 1 kHz			
Input range (deg/s)	NA	± 375	± 90	± 150
Size (mm)	190.5 x 114.3 x 45.7	88.9 x 58.4 x 33.0	8 x 16 x 5	7 x 7 x 3.5
Weight (g)	1360	270	1	<1

2.3.3. Inertial navigation systems with IMU

In the area of inertial navigation systems (INS), estimating pose of a moving object using gyroscopes and accelerometers placed on the object is a well-introduced concept. Generally, there are two approaches of INS designs.

- (i) gimbaled or stabilized platform approach, and
- (ii) strapdown approach.

Gimbaled or stabilized platform approach is employed in earlier navigation systems in which gimbals or pivoted-frames are attached to a so-called “stabilized platform” onto which gyroscopes and accelerometers are placed as shown in Fig. 2.5. The orientation of the platform is made stationary to a particular reference frame (e.g., world frame) by means of actuators which are usually servo-motors attached to the gimbals using angular velocity feedback information from gyroscopes. Since orientation of the platform with respect to the reference frame does not change, acceleration outputs from the accelerometers can be double-integrated to obtain positions of the moving object in the respective accelerometer sensing directions. The platform approach is still employed in some applications requiring very accurate estimates of navigation data, such as ships and submarines.

Strapdown approach is employed in modern INSs in which three gyroscopes which are orthogonal to each other are fixed or “strapped down” to the moving object as shown in Fig. 2.6. Three orthogonal accelerometers are also fixed to the object. The mechanical complexity involved in the platform approach is eliminated in the strapdown approach.

Instantaneous orientation of the moving object is calculated by integrating the angular velocity information obtained from the gyroscopes using initial orientation information. Instantaneous position of the moving object is obtained by double-integration of acceleration in world frame using initial position information. The acceleration in world frame or inertial frame is obtained from the acceleration in body frame by transformation using the known instantaneous orientation of the body determined by the gyros. Block diagram of a modern INS with accelerometers and gyroscopes are shown in Fig. 2.7.

Fig. 2.5. An INS with a stabilized platform approach [Foxlin 02]. Accelerometers and gyros are fixed on the stabilized platform.

Fig. 2.6. A strapdown INS [Foxlin 02]. Accelerometers and gyros are fixed directly to the mobile object.

Fig. 2.7. Block diagram of the flow of obtaining instantaneous position of a moving object in strapdown approach INS.

2.3.4. Gyroscope-free INS

Accelerometers are employed in most inertial navigation systems to sense linear accelerations and gyroscopes to sense angular velocity. An accelerometer-based or gyroscope-free inertial navigation system (GF-INS) that uses only accelerometers to compute the linear and angular motions of a rigid body has been introduced for a few decades. The main motivation for developing a gyroscope-free sensing system is that low-cost gyroscopes lack the accuracy needed for precise navigation application.

In theory, a minimum of six accelerometers are required for a complete description of rigid body motion. An original cube configuration design with six accelerometers, the minimum number of sensors to recover all the six kinematic parameters, is proposed by Chen *et al.* [Chen 94]. Nine-accelerometer scheme is proposed by Wang *et al.* [Wang 03] to get angular velocity directly without integration from angular acceleration. Tan and Park [Tan 05] analyze the effects of accelerometer location and orientation errors on the angular and linear motions calculation result. It is shown by Ang *et al.* [Ang 03] that sensing resolution of angular motion derived from accelerometers placed inside Micron [Riviere 03] is higher than that obtained from the equivalent grade gyros. Consequently, gyroscope-free IMU is employed in the sensing sub-system of Micron [Ang 03].

2.4. Hybrid Sensing Systems

Tracking systems that employ only one sensing modality all have to experience intrinsic disadvantages. For example, sensing system which employs only inertial sensors suffer from drift, optical tracking systems require a clear line-of-sight, and magnetic tracking systems are affected by ferromagnetic and conductive materials in the environment [Raab 79]. Hybrid tracking systems have become popular since they can eliminate the intrinsic disadvantages.

For example, Foxlin has developed a system that is primarily inertial, but aided by ultrasonic sensors [Foxlin93]. An example of such system is IS-600 Acousto-Inertial Hybrid 6-DOF Tracking System which is shown in Fig. 2.8. The system tracks changes in orientation and position by integrating the outputs of its gyros and accelerometers, and corrects drift using a room-referenced ultrasonic time-of-flight range measuring

system. It consists of an I-Cube IMU, SoniDisc ultrasound transmitter, and a ReceiverPods ultrasounds receiver.

Fig. 2.8. Schematic diagram of IS-600 Acousto-Inertial 6-DOF Tracking System [Foxlin 98a]. The tracking system is a hybrid system consisting of internally-referenced inertial tracking and externally-referenced ultrasonic tracking.

The ReceiverPods placed at known locations broadcast an infrared trigger code which uniquely identifies one of up to 8 differently coded SoniDiscs. The selected disc responds immediately with a 40 KHz ultrasonic pulse. Separate timer counters are started in each of the ReceiverPods at the instance of the IR broadcast, and halted by the arrival of the pulse at the ReceiverPod microphone. Using the value of the speed of sound (which is calculated from the measured temperature), range measurements are obtained at the three ReceiverPods, and used to compute position. Kalman filtering is used to fuse two pieces of information from I-Cube and ultrasonic range measurement.

Canadian Aerospace Electronics [NRC 94] and the University of North Carolina [Meyer 92] have pursued hybrid systems that are primarily optical, but aided by inertial sensors (for prediction), and researchers at the University of Tokyo have sought to improve the data rate of the Polhemus tracker by augmenting it with rate gyroscopes [Emura 94], while researchers at the University of North Carolina have sought to improve the accuracy of the Ascension magnetic tracker by augmenting it with a passive image-based system that observes known fiduciary marks in the real world [State 96].

Sensing system of Micron can be considered as hybrid. Main sensing elements in Micron are accelerometers while a magnetometer served as an aided sensor to correct for the drift.

2.5. Choice of Sensors for *ITrem* Sensing Sub-system

As can be seen from the earlier sections, since the sensing requirements stated on page 13 in section 2.1 can be fulfilled by none of the commercially available externally referenced sensing systems and it is shown in [Ang 03] by Ang *et al.* that sensing resolution of angular motion obtained from micromachined accelerometers is better than that from micromachined gyros, the sensing sub-system will be developed using accelerometers. Although a magnetometer or a similar aided sensor is usually required to correct for the drift caused by accelerometers, with the new algorithm which will be presented in Chapter 4, no aided sensor such as a magnetometer is required to detect the tremulous motion. ADXL-203 dual-axis accelerometers from Analog Devices [Analog 08] will be employed since they are still the best choice accelerometers in terms of sensitivity, weight, and size among the other accelerometers listed in Table 2.2.

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Chapter 3 : Optical Micro Motion Sensing System (M^2S^2)

The chapter describes the needs to develop a micro motion sensing system, the system design and development, calibration of the system using neural network, elimination of light disturbances, and system performance.

3.1. Introduction

In this section, the motives for design, development, and calibration of an optical Micro Motion Sensing System called M^2S^2 are described. As will be seen in the subsequent chapters, the design and development of the sensing sub-system of the tremor compensation instrument involves a number of tests and evaluations. These include validation of the designs of placement of accelerometers for high angular sensing, and evaluation of the overall performance of the sensing sub-system.

Evaluation of the performance of the sensing sub-system includes comparing instrument tip positions estimated/calculated by the sensing sub-system of the instrument with the actual tip positions. Evaluation of the performance of the manipulation sub-system includes comparing the desired or commanded positions of the tool tip with the actual tip positions. Likewise, evaluation of the tremor compensation performance of the instrument, which is the combination of all the sub-systems' performances, includes the comparison between uncompensated and compensated tool tip oscillations. In all the evaluations, the tip positions need to be measured accurately with a sensing system apart from the sensing sub-system of the instrument.

A particular sub-system (e.g., sensing sub-system) of the instrument has to be tuned, modified, or redesigned until the evaluation results are satisfactory. Therefore, performance evaluation of each sub-system of the instrument is one of the unavoidable steps towards the successful development of the tremor compensation instrument. Accurate evaluation requires accurate measurement of the tool tip. Hence, obviously, a sensing system which has better specifications than those presented in section 2.1 on page 13 but does not necessarily possess portability, and non-obtrusiveness features is essentially required for the development of the tremor compensation instrument.

Another good reason for developing the M^2S^2 is as follows. As described in 0, many involuntary and inadvertent components are present in normal human hand movement. These include physiological tremor [Harwell 83], jerk [Schenker 95], and

low frequency drift [Hotraphinyo 01]. These undesirable components have limited the manipulation accuracy of surgeons in microsurgery, and cause certain types of procedures to be generally infeasible, such as retinal vein cannulation and arteriovenous sheathotomy [Tang 00]. To better engineer and evaluate performance of manipulation accuracy enhancement systems such as those described in section 1.2, better knowledge of these components present in surgeons' hand movement during microsurgical operations is required.

Several one-dimensional (1-D) studies of motion have been reported for microsurgeons [Harwell 83; Riviere 97] and medical students [Walsh 94]. Some studies have examined only physiological tremor [Harwell 83; Walsh 94], the roughly sinusoidal component of involuntary motion. Others have examined also the general motion of surgeons [Hotraphinyo 01; Riviere 97]. Studies of the motion of microsurgeons in 3-D are rare in the literature. That is partly because of the fact that commercially available motion tracking systems described in Chapter 2 such as Optotrak (Northern Digital, Waterloo, Canada) [Simon 97], Isotrak II (Polhemus, Colchester, Vt.), miniBIRD (Ascension Technology Corp., Burlington, Vt.) [Tseng 99], Fastrak (Polhemus, Colchester, Vt.) [Sherman 99], and the HiBall Tracking System [Welch 01] cannot provide adequate accuracy and resolution for micro scale motion.

To address the above issues, Riviere *et al.* developed ASAP (Apparatus to Sense Accuracy of Position) and MADRID (Measurement Apparatus to Distinguish Rotational and Irrotational Displacement) [Hotraphinyo 01; Riviere 99; Ortiz 03] systems to track microsurgical instrument tip motion in micro-scale. Position Sensitive Detectors (PSDs) and lenses form the main components of these systems whereby the accuracy is limited by nonlinearity inherent in PSDs and lenses. However, these systems are not commercially available and are not accessible to the author. This necessitates the author developing a sensing system for the development of the tremor compensation instrument.

It is well known that lens introduces distortion. PSDs also introduce distortion effects that vary from PSD to PSD and these distortion effects play a significant part at submicron and nanometer levels. The overall nonlinearity is the combination of distortion effects from the lenses and PSDs. Nonlinearity and ambient light effects are not taken into account in systems described in [Hotraphinyo 01; Riviere 99; Ortiz 03]. The accuracies of the systems at different tilt orientations of the instrument are not

investigated nor described although they are different at different orientation due to imperfection in absorption of IR lights by the non-reflective instrument shaft. Accuracy is worse without calibration because of imperfect alignment and assembly of the components. If accuracy and resolution can be improved by eliminating inherent nonlinearity and increasing signal to noise ratio respectively, the optical Micro Motion Sensing System (M^2S^2) can be used as a perfect ground truth to evaluate accuracy enhancement devices [Das 99; Taylor 99; Ang 01] and an accurate system for the assessment and microsurgical trainings of surgeons. This motivates the author to develop a better tracking system (M^2S^2) and calibration method to eliminate nonlinearity inherently present in the system.

3.2. Methodology

3.2.1. System design and development

The system consists of two PSD modules, three lenses, an IR diode, a white reflective ball and a data acquisition (DAQ) system as shown in Fig. 3.1 and Fig. 3.2. The overall system block diagram is shown in Fig. 3.3.

Fig. 3.1. Schematic Diagram (Top View) of the system.

Fig. 3.2. Micro Motion Sensing System (M^2S^2) and the location of the workspace shown in dotted line

Instead of expensive high resolution and speed CCD or CMOS cameras being used, PSD cameras are proposed to be used as PSDs offer adequate high position resolution and fast response [Schaefer 98] at a lower and affordable cost. While CCD or CMOS camera's resolution is limited by the number of pixels, that of PSD is limited by the noise floor of the system.

Fig. 3.3. Overall block diagram of the system (M^2S^2)

A white reflective ball (Diameter=6mm) from Gutermann (Art. 773891), which is to be tracked, is attached to a 1 mm diameter rod or instrument shaft painted with a non-reflective black color. IR light is shone onto the workspace containing the ball so that IR light reflected by the ball would impinge on the PSDs. The light spot positions on the respective PSDs are affected by the ball position.

Bi-convex lenses (Diameter=25.4mm, f=25.4mm) from Thorlabs (LB1761-B) are used to focus the reflected IR light onto the PSDs. These anti-reflection lenses also reduce surface reflection to maximize the IR light intensity striking the PSDs. Without

the bi-convex lenses, the reflected IR light would impinge on the PSDs in such a manner that the position of the ball position cannot be determined correctly.

Reflection from the Ball

Not all the IR lights that hit the ball enter the lenses. Only some portion of the lights passes through the lenses and falls onto the PSD surface. Plan view schematic diagram of reflection of IR lights from the ball surface is shown in Fig. 3.4. Fig. 3.4 (b) shows close-up view of incidence of IR lights onto the ball. The surfaces that reflect the lights which enter the lenses are also shown in the figure.

Fig. 3.4. Plan view schematic diagram of reflection of IR lights from the ball surface.

Location and Shape of Workspace

Shape and location of the workspace with respect to the lenses is shown in Fig. 3.5. Close-up view schematic diagram of the workspace is shown in Fig. 3.6. The region enclosed with blue lines (and filled with dashed lines) represents the workspace within which the ball can be seen by both cameras.

As can be seen from the geometry, magnification is not constant within the workspace. As can be seen from Fig. 3.6, the maximum magnification of 1.4 occurs at location 1 while the minimum magnification of 0.8 occurs at location 2. The

magnification is one at the intersection point of the two principal axes of the cameras. A 3D view of the workspace together with dimensions is shown in Fig. 3.7.

Fig. 3.5. Top view schematic diagram of shape and location of the workspace with respect to the lenses.

Fig. 3.6. Close-up view schematic diagram of the workspace.

Fig. 3.7. A 3D view of the workspace.

The PSD modules

Two PSD modules (DL400-7PCBA3) from Pacific Silicon Sensor Inc are used in the system and placed orthogonally to each other (Fig. 3.1 and Fig. 3.2). Each module employs a two-dimensional PSD with a 20mm square sensing area. There are two pairs of output electrodes in a two-dimensional PSD – X-axis electrode pair and Y-axis electrode pair. The current produced at an electrode depends on the intensity and position of the light spot centroid on the PSD. With regards to an electrode pair, the electrode located nearer to the light spot centroid will produce more current than the other. With this phenomenon, the position of the light spot centroid can be calculated once the current produced at each electrode is known. The module converts current produced at each electrode into voltage and outputs difference and sum of the converted voltages of each electrode pair: V_{Xdiff} and V_{Xsum} for X-axis electrode pair and V_{Ydiff} and V_{Ysum} for Y-axis electrode pair.

Making calculated position independent of light intensity

The difference outputs (V_{Xdiff} and V_{Ydiff}) can be used to calculate X and Y coordinate positions of the light centroid. Calculated positions are susceptible to fluctuations in light intensity. Therefore, the sum outputs (V_{Xsum} and V_{Ysum}) are used to normalize the difference outputs so that the X and Y positions become independent of fluctuations in light spot intensity. The position calculation is shown in the equations below.

$$P_X = \frac{V_{Xdiff}}{V_{Xsum}} = \frac{V_{X1} - V_{X2}}{V_{X1} + V_{X2}} \quad (3.1)$$

$$P_Y = \frac{V_{Ydiff}}{V_{Ysum}} = \frac{V_{Y1} - V_{Y2}}{V_{Y1} + V_{Y2}} \quad (3.2)$$

Where P_X and P_Y are normalized centroid position of the fallen light on the PSD surface in X and Y axis respectively. V_{Di} , $D = X, Y; i = 1, 2$, is converted voltage from the current obtained at electrode i of D electrode pair.

(a)

(b)

Fig. 3.8. Block diagram of (a) how normalized positions are obtained from the lights falling on a PSD, and (b) how X, Y, and Z position of the instrument tip is obtained from normalized positions of light spots on both PSDs.

The normalized position values from both PSDs, $P_D^k, k = 1, 2$, obtained are then used as inputs to the neural network to get X, Y, and Z position of the instrument tip. The block diagram of how X, Y, and Z position of the tip is obtained from the lights falling on the PSDs is shown in Fig. 3.9.

Data acquisition and increasing signal-to-noise ratio

Voltage signals from each output of the PSD modules are sampled and digitized at 16.67 kHz, 16 bits using the PD2-MF-150 data acquisition card from United

Electronic Industries, Inc. Shielded cables are used to connect every PSD module output to the data acquisition card for better noise suppression. The power and signal cables are also separated to eliminate coupling of power line noise and signals. The digitized samples are first filtered with an 8th order Butterworth low-pass software filter having a cutoff frequency of 50Hz and subsequently average every 67 samples. Thus, the overall sampling rate is $16.67\text{k}/67=248\text{Hz}$. The unfiltered and filtered signal noise is about 10mV and less than 0.5mV respectively. After averaging, the filtered voltage signals V_{Ddiff} and V_{Dsum} are used to compute the respective normalized positions $P_D^k, k=1,2$.

The system resolution is determined by the signal to noise ratio (SNR). A system with higher SNR would offer better resolution. The intensity and wavelength of reflected IR light striking the PSDs are important factors in increasing SNR. The intensity must be high enough whilst the IR light wavelength must fall within the most sensitive region of the PSD spectral response. OD-669 IR diode from Opto Diode Corp. is chosen as the IR light source because it has the highest power output and its peak emission wavelength is within the most sensitive region of the PSD spectral response. Since radiation of IR light from OD-669 is diverging, an aspheric condenser lens ($D=27\text{mm}$, $EFL=13\text{mm}$) from Edmund Optics is placed in front of the IR diode to converge the IR rays onto the workspace. The lens also enhances the IR intensity by more than 2 times. The IR diode is pulsed at 500Hz (Duty cycle=50%) to increase the maximum current limit of the IR diode as well as to handle undesired lights. Increasing maximum current limit results in a higher IR emission intensity and thereby, improving the SNR. The pulse timing of the IR diode is controlled with a PIC microcontroller via a field effect transistor (FET).

3.2.2. System calibration

Calibration of the system can be performed by various approaches such as physical modeling, brute force method in which look-up table (LUT) and interpolation are used, neural network, etc. Physical modeling can give the best accuracy if all the parameters of the system are taken into consideration but that requires thorough understanding of the physical behavior of each individual component. LUT with interpolation method requires calibration data points be stored in memory and needs more data points for higher accuracy.

A multilayer neural network approach is proposed for this application because it eliminates the need to know individual component's behavior; uses a reasonable span of time for a trained network to determine the output and can approximate any arbitrary continuous function to any desired degree of accuracy [Funahashi 89].

The calibration data points are obtained by moving the ball being tracked to known locations and using the corresponding output values from the two PSDs. Movement of the ball is automatically done by controlling motorized precision stages using software developed in-house. X-Y-Z stepper-motorized precision translation stages (8MT173-20 from Standa) are used for the calibration. According to the specifications, 1 step movement of the stepper motors is equivalent to 1.8 degrees rotational and 1.25 micrometer translational movement respectively. The step angle accuracy is 5%. Since the data points spacing is 1000 micrometers which is an integer multiple of 1.25 micrometers, the accuracy is $1.25 \times 0.05 = 62$ nanometers. The travel range of the stages is 20 mm. The calibration setup for the system is shown in Fig. 3.9.

The center of the M^2S^2 workspace and the center of the 20 mm³ space of travel provided by translation stages do not coincide. Out of 20 mm³ provided by the translation stages, only approximately 12 mm³ overlaps with the workspace. This is mainly due to limited choices in positioning of the translation stages with respect to the M^2S^2 system. Therefore, only the calibration data within the 12 mm³ is used for the neural network training and evaluation of the system accuracy. The spacing between two successive calibration points is 1 mm. The total number of calibration data points for 12 mm³ is therefore $13 \times 13 \times 13 = 2197$. The normalized positions obtained at each calibration point and their corresponding absolute positions are used as training and target values respectively for neural network training.

In order to improve the quality of the acquired data, after every step movement, sufficient time is provided for mechanical vibrations generated by the motion of the motorized stages to settle down before acquiring data. In addition, the system including the motorized stages is placed on an anti-vibration table to suppress floor emanated vibrations.

In order to ensure that calibration is not affected by temperature, the system is switched on for about 50 minutes before commencing calibration so that the IR diode attained a steady operating temperature state.

Fig. 3.9. M^2S^2 system and its calibration setup.

Mapping by a feedforward neural network

A multilayer artificial neural network model is proposed because PSD calibration is a nonlinear problem that cannot be solved using a single layer network [Fausett 94]. The determination of proper network architecture for the PSD system calibration is ambiguous like other neural network applications. The best architecture and algorithm for a particular problem can only be evaluated by experimentation and there are no fixed rules to determine the ideal network model for a problem. However, variations in architecture and algorithm affect only the convergence time of the solution.

The network model employed includes 4 input neurons, and 3 output neurons. The number of neurons in the hidden layers and the number of layers are varied to achieve the best accuracy. Four neurons are used in the input layer because there are

four inputs: P_X^1, P_X^2, P_Y^1 , and P_Y^2 . Three neurons are used in the output layer for the three coordinate positions: X, Y , and Z . Transfer functions of all the layers are log-sigmoids except for the output layer, which is a linear function.

3.2.3. Handling ambient and environment lights

Total lights falling on the PSDs, L_{tot} , comprise three components: reflected IR light from the ball L_{IRball} , reflected IR light from the environment (surrounding objects) L_{IRenv} , and ambient light L_{amb} .

$$L_{tot} = L_{IRball} + L_{IRenv} + L_{amb} \quad (3.3)$$

Since the instrument tip position, which is the same as the ball position, is of interest, only the voltage outputs due to L_{IRball} should be used in the calculation of normalized positions P_D^k , and voltage outputs due to the other two components can be regarded as DC offsets or disturbances and they should be excluded in the calculation P_D^k .

Effect of different ambient light condition

The system described so far does not account for the effect of different ambient light conditions and hence ambient light has to be maintained the same during and after the calibration data are obtained. However, when the system is placed under a microscope for the assessment of micromanipulation accuracy, ambient light is different from the one that exists during the calibration period and measurement accuracy of the system is affected. The effect is shown in Fig. 3.10.

Fig. 3.10. The effect of changes in ambient light on the accuracy of M^2S^2 .

For a stationary ball, position values calculated by M^2S^2 should be constant regardless of the ambient light condition, but it gives different position values at different light conditions.

To see the effect of different ambient light intensity on the accuracy of the M^2S^2 , a light bulb (40W, 240Hz) is placed on the system and turned on and off a few times. As can be seen in the figure, the calculated position of the ball is not the same for the periods while the light is turned on (Light ON) and off (Light OFF).

Handling ambient light

As mentioned earlier, the IR diode is pulsed at 500 Hz (Duty cycle = 50%). The lights falling on the PSDs during the period while the IR diode is switched off, L_{Off} is due only to ambient light, L_{amb} . The lights falling on the PSDs during the period while the IR diode is switched on, L_{On} normally consist of all three components described in (3.3).

$$L_{On} = L_{tot} \quad (3.4)$$

$$L_{Off} = L_{amb} \quad (3.5)$$

Subtracting (3.5) from (3.4) and using (3.3), resultant light, L_{res} becomes

$$\begin{aligned} L_{res} &= L_{On} - L_{Off} = L_{tot} - L_{amb} \\ &= L_{IRball} + L_{IRenv} \end{aligned} \quad (3.6)$$

Fig. 3.11. The calculated X, Y, and Z positions of the instrument tip invariant to different ambient light intensities.

By subtracting the voltage outputs obtained during the period while the IR diode is switched off, V_{Off} from those obtained while it is on, V_{On} , the voltage outputs of the module due to L_{res} are obtained.

The normalized position in D axis of the light spot on PSD k obtained from the voltage outputs due to L_{res} is symbolized as ${}^{res}P_D^k$ which are used as input to the network. The position output becomes independent of the ambient light condition and the effect of changes in ambient light is eliminated as shown in Fig. 3.11.

Neural network has to be trained previously with the input data consisting of ${}^{res}P_D^k$ obtained from L_{res} (the combination of L_{IRball} and L_{IRenv} only), and not obtained from L_{tot} . When the system is run, L_{Off} (or V_{Off}) and L_{On} (or V_{On}) are calculated on-line and subtraction of the two values is made to get voltage outputs due

to resultant light L_{res} . ${}^{res}P_D^k$ obtained from L_{res} are then used as the input to the neural network which gives the position output which is invariant to ambient light condition.

Handling changes in surrounding objects

Since the resultant light still includes IR light from the surrounding objects (the surrounding objects which are under the exposure of IR light) even after the subtraction, they have to be maintained the same. If the properties such as shape, color, reflection coefficients, etc of surrounding objects are changed or extra objects are added or removed for some reasons, L_{IRenv} changes and subsequently neural network gives inaccurate position outputs. Hence, the network has to be trained again with normalized positions obtained from voltage outputs caused by the combination of L_{IRball} and new L_{IRenv} (i.e., IR light from the environment after the changes in surrounding objects' properties). To do so, calibration has to be carried out again. Performing calibration requires some preparation and takes several hours in getting the whole calibration data set.

In order to eliminate the need to do calibration again in the case surrounding objects or their properties are changed, voltage outputs due to L_{IRenv} should also be excluded and those due to L_{IRball} only should be used to get input to the network. To use the voltage outputs due to L_{IRball} only, those due to L_{IRenv} need to be known. It is known by carrying out the following steps.

The ball which is attached to the instrument tip is removed from the workspace and IR diode is switched on and off. Then, L_{On} contains only two components; L_{IRenv} and L_{amb} . By subtracting voltage outputs due to L_{Off} from those due to L_{On} , voltage outputs due to L_{IRenv} can be obtained.

$$L_{IRenv} = L_{On} - L_{Off} \quad (3.7)$$

Voltage outputs due to L_{IRenv} remains constant as long as surrounding objects are maintained the same. Whenever the surrounding objects or their properties are changed, the above steps can be easily carried out to find voltage outputs due to the new L_{IRenv} .

To prove the invariance of system accuracy by the changes in environment (surrounding objects or their properties), a test platform (a kind of objects) is placed under the workspace of M^2S^2 as shown in Fig. 3.12. The presence of the test platform makes L_{IRenv} different from the time the system is calibrated (i.e., without the presence of the platform).

Two neural networks are used for comparison. The first neural network is trained with normalized positions ${}^{res}P_D^k$ obtained from voltages due to the lights which consist of a combination of L_{IRball} and L_{IRenv} without the presence of the platform (as in Fig. 3.2). Therefore, when the test platform is present (as in Fig. 3.12), the accuracy is affected to some extent depending on the difference in L_{IRenv} between with and without the presence of the platform.

Fig. 3.12. M^2S^2 with test platform.

The input training data of the second neural network is the normalized positions ${}^{IRball}P_D^k$ obtained from voltages due to the light which consists of L_{IRball} only. Therefore, accuracy is not affected by the presence of the test platform. Table 3.1 shows

the comparison of Root Mean Square Errors (RMSE), maximum errors, and accuracies of the system for the two approaches.

Table 3.1 Comparison of *RMSE*, maximum errors, and accuracies of the system for the two approaches

The data with which neural network is trained	$IR_{ball} P_D^k$	$res P_D^k$
RMSE (μm)	83.44	115.96
RMS Accuracy (%)	99.3	99.03
Maximum Error (μm)	224.4	430.00
Minimum Accuracy (%)	98.13	96.42

The RMSE in percentage (RMSE (%) and Maximum Error (%)) are calculated over the range of 12 mm cubical space. To determine the measurement errors at each point, measured values at these locations are subtracted from their respective true values. True values are known by moving the ball to be tracked to known locations using motorized precision translation stages. Then, Root Mean Square (RMS) error and maximum error values are calculated using errors at all the points. *RMS* accuracy and minimum accuracy are calculated as follows.

$$\text{RMS accuracy (\%)} = \frac{12000 - \text{RMSE}}{12000} \times 100(\%) \tag{3.8}$$

$$\text{Minimum accuracy (\%)} = \frac{12000 - \text{MaxErr}}{12000} \times 100(\%) \tag{3.9}$$

where *MaxErr* is maximum error in microns.

Transient accuracy

The transient behavior of the system when ambient light is changed is shown in Fig. 3.13. The system accuracy is affected only during the period of about one second after the ambient light intensity is changed. The effect is discussed in section 3.6.

Fig. 3.13. Close-up view of the transient effect when the light is switched off.

3.3. Software Implementation and Integration

LabVIEW, a signal acquisition, measurement and analysis software from National Instrument Corporation, is used to acquire, filter, and process signals from the PSDs. Neural network function is implemented in C language and integrated within the LabVIEW based program developed.

Since acquisition takes place continuously while the IR diode is being pulsed, the acquired signals contain voltage values whilst the IR diode is switched on (ON-time) and off (OFF-time). Therefore, acquired voltage values need to be segregated for V_{On} and V_{Off} . The segregation is carried out using a software threshold approach described hereafter.

In this approach, a transistor-transistor logic (TTL) signal having the same switching frequency and phase as the signal controlling the IR diode is used as the reference signal. TTL is about 5V and 0V during ON-time and OFF-time respectively. A data acquisition channel is used to acquire the TTL signal and subsequently,

compared with predefined threshold values of 4V and 0.05V for V_{On} and V_{Off} respectively.

One scan is defined as a period during which sampling takes place channel after channel for all active data acquisition channels. In a scan, if the acquired TTL value is above the ON-time threshold, all voltage values acquired during that scan are regarded as V_{On} . Similarly, if the TTL value is below the OFF-time threshold, all acquired voltage values are considered as V_{Off} .

3.4. Resolution and Accuracies of the System

Resolution of the system is determined by placing a stationary ball in the middle of the workspace and measuring the ball position for a period of one minute. Resolution or standard deviation (STD) of the measured values during one minute is found to be $0.9\ \mu\text{m}$.

When the instrument is in use, its orientation might be different from the one at which M2S2 is calibrated. It is important to know how much accuracy is out at a particular orientation of the instrument. Therefore, after the system is calibrated with the instrument orientation being vertical, the accuracy is tested at different orientation angles (Fig. 3.14). Different accuracy results at different instrument orientation angles are reported in Table 3.2.

A 3D plot of measurement accuracy of M^2S^2 for the instrument oriented vertically is shown in Fig. 3.15. A 3D plot of measurement accuracy of M^2S^2 for the instrument slanted 60 deg from the vertical axis is shown in Fig. 3.16.

Fig. 3.14. Measuring ball positions at known locations with the instrument being slanted at 45 degrees from the vertical axis for accuracy testing.

Table 3.2 *RMSE* and maximum errors of the system at different orientation of the instrument

(a) No slanting

RMSE (μm)	83.436
RMS Accuracy (%)	99.3
Maximum Error (μm)	224.4
Minimum Accuracy (%)	98.13

(b) 30 deg slanting

RMSE (μm)	211.06
RMS Accuracy (%)	98.24
Maximum Error (μm)	586.77
Minimum Accuracy (%)	94.57

(c) 45 deg slanting

RMSE (μm)	435.79
RMS Accuracy (%)	96.37
Maximum Error (μm)	1654.32
Minimum Accuracy (%)	86.21

(d) 60 deg slanting

RMSE (μm)	501.07
RMS Accuracy (%)	95.82
Maximum Error (μm)	1854.12
Minimum Accuracy (%)	84.55

Fig. 3.15. Three dimensional plot of measurement accuracy of M^2S^2 for the instrument oriented vertically as shown in Fig. 3.9.

Fig. 3.16. Three dimensional plot of measurement accuracy of M^2S^2 for the instrument slanted 60 degrees from the vertical axis.

3.5. Dynamic Performance of the System

The accuracy values reported in Table 3.1 and Table 3.2 are static accuracy values which are calculated from measurement values which are obtained by moving the ball to known locations and measuring the respective positions of the ball when it is stationary (i.e, measurement is not done when the ball is being moved).

Dynamic accuracy of the system might be different from static accuracy due to the filter employed in the system and some other factors. Therefore, dynamic accuracies of the system at different frequencies are measured. Dynamic accuracies are obtained by measuring position of a ball moving in a sinusoidal motion with 50 μm amplitude with varying frequencies up to 12Hz. Movement of the ball in a sinusoidal motion is generated by a nanopositioning system (P-561.3CD from Physik Instrumente). The positioning accuracy of the system is in terms of submicron. A plot of positioning accuracy of nanopositioning system for a sinusoidal motion of 100 μm p-p amplitude at 10 Hz with an offset value of 50 μm is shown in Fig. 3.17.

Fig. 3.17. A plot showing positioning accuracy of nanopositioning system for a one-cycle sinusoidal motion of 100 μm p-p amplitude at 10 Hz with an offset value of 50 μm .

Measured positions by the system and ground truth or known positions which are generated by the nanopositioning system are compared. Dynamic accuracies are shown in Table 3.3. The plot of the displacement of the ball at 12 Hz and its displacement error are shown in Fig. 3.18.

Table 3.3 Static and dynamic errors

	@ 8 Hz	@ 12 Hz
Dynamic RMSE (μm)	1.2	1
Dynamic RMS Accuracy (%)	98.8	99

Dynamic RMS accuracy is calculated based on the measurement range of 100 μm as follows.

$$\text{Dynamic RMS Accuracy (\%)} = \frac{100 - \text{RMSE}}{100} \times 100 (\%)$$

(a) displacement

Fig. 3.18. (a) displacement, and (b) displacement error.

3.6. Discussion

When using normalized positions (${}^{IRball}P_D^k$) obtained from voltage outputs due to IR light from the ball only, as input to the neural network, transient accuracy of the system is affected by the changes in ambient light only for about one second. After about one second, the system can provide position information accurately. The transient inaccuracy is believed to be due to the low pass filter employed. However, in most situations, ambient light intensity is not changing during the usage of the system. For example, when it is used under a microscope, the microscope light is fixed at a particular intensity after the user has adjusted it to suit him or her. The method can provide accurate position measurement without the need to have the same ambient light condition as the one that exists during calibration.

The accuracy is getting lower at higher slanting angles of the instrument. The inaccuracy is introduced by the imperfection in absorption of IR light by non-reflective paint on the instrument shaft. There is some reflection of IR light from the instrument shaft. The patterns of the reflection are different at different orientations of the instrument. At higher slanting angles, the difference in patterns of reflection from the shaft between the vertical orientation and the orientation at the slanting angles is higher and hence the accuracy is lower.

Therefore, to get higher accuracy, the instrument orientation should be as vertical as possible because normally, the calibration is carried out with the vertical

(downward) orientation of the instrument. In the case that manipulation tasks involving the instrument slanting at a particular angle is required, the system should be calibrated with the instrument orientated at that particular angle for maximum accuracy.

The dynamic RMS accuracy shown in Table 3.3 is almost the same as static accuracy shown in Table 3.2 (a). The small amount of difference is possibly due to the movement errors of the positioning device. No attenuation of the signal is found at tested frequencies. Testing the system's dynamic accuracies within only 8 to 12 Hz is enough because physiological tremor frequency of human hand is in the range of about 8 to 12 Hz.

3.7. Conclusion

One of the contributions in this chapter come from the fact that the system is an essential assistive tool for the development of *ITrem* as well as an evaluation tool to assess the compensation performance of *ITrem*. Measurement of the system is invariant to ambient light changes, and hence can be used at any places such as under different microscopes.

Another contribution is that, as described in the introduction of this chapter, it can be used to evaluate the performance of accuracy enhancement systems and the manipulation performance of operators in micromanipulation tasks such as microsurgeries. Moreover, it opens up the research area to better study the components of hand movement present in various microsurgical operations and conditions. For example, the study reported in [Su 09] is a consequence the development of M^2S^2 .

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Chapter 4 : Physiological Tremor Sensing using only Accelerometers for Real-time Compensation

This purpose of this chapter is to show that accelerometers only are good enough in obtaining tremulous motion of the instrument tool tip. The chapter introduces the concept of “Steady-tip frame” for describing tremulous motion of the tip with respect to the non-tremulous motion. An algorithm which consists of zero-phase real-time band-pass filtering and makes use of acceleration readings from accelerometers to provide the tip tremulous motion required for compensation is described. As part of the algorithm, filtering low-frequency motion components without introducing discernible phase-lag to be suitable for the zero-phase filtering, and kinematics to obtain tip motion from the body motion are described. The algorithm is proved with experimental data.

Rigid-body assumption which is valid in most situations, in which the motion of the instrument and that of the surgical tool attached to the instrument is assumed to be the same, is used in inferring the tip motion from the accelerometer outputs. Rigid-body assumption is valid in most micromanipulation especially in vitreoretinal microsurgery since the forces required to exert on the tissues by a surgical tool is extremely small; less than 7.5 mN in magnitude most of the time [Gupta 99] and it is also concluded that the surgeon cannot probably feel the interaction between the surgical tool and retinal tissue most of the time.

At the end of the chapter, advantages obtained due to the algorithm which the author proposed is summarized.

4.1. Introduction

Sensing sub-system of tremor compensation device proposed by Bose *et al.* [Bose 92] is an optical-based sensing system similar to M^2S^2 . Being an optical-based system, line-of-sight between external sensors and the surgical tool has to be maintained causing intrusiveness to the surgeon. Moreover, unwanted light reflection from the objects other than the surgical tool tip causes inaccuracy of the sensing system.

Sensing sub-system of Micron comprises three dual-axis accelerometers and a magnetometer [Ang 04]. Schematic diagram of locations of the sensors in Micron is

shown in Fig. 4.1. The use of the magnetometer is to provide complementary pan angle information for obtaining the information on complete orientation of the instrument.

There are weaknesses in the sensing sub-system of Micron. One of the weaknesses is requirement of on-site calibration of the magnetometer [Benjavitvilai 05]. If the environment magnetic field is changing with time, for example, due to intermittent usage of equipments which generate electromagnetic fields, calibration data cannot be employed.

Another weakness is related to the estimation of tip motion. Estimate of complete orientation of the instrument is obtained by fusing roll-angle and tilt-angle information from accelerometers and pan angle information from the magnetometer via Kalman filtering and it is stated that estimate of the instrument orientation is sub-optimal [Ang 04]. Therefore, estimation of the motion of the tip must be sub-optimal since kinematic calculation of the tip motion requires the estimate of instrument orientation. The need of the complete orientation information and hence the need of a magnetometer results from the approach to obtain the information on instrument tool tip motion with respect to a world frame.

If the tool tip motion can be estimated without using a magnetometer, the issues mentioned above can be eliminated. Therefore, it is worth exploring alternative methods/approaches or sensing modalities without using a magnetometer. As far as the compensation for the tool tip motion due to physiological tremor is concerned, the information on the tool tip motion with respect to the world frame is not required. As long as the tool tip motion due to physiological tremor with respect to the tool tip motion due to the other components rather than the physiological tremor is known, compensation can be carried out. Based on this concept, an approach which eliminates the need to use a magnetometer to sense the motion of the tool tip is proposed. In this chapter, feasibility of the proposed approach of physiological tremor sensing using only accelerometers is presented.

Fig. 4.1. Schematic diagram of locations of sensors in Micron [Ang 04].

4.2. Methodology

4.2.1. Tremulous motion and steady tip frame

Motion sensed by the accelerometers fixed on the instrument body consists of tremulous motion which results from physiological tremor, and non-tremulous motion

which results from non-tremulous motion of the hand; intended motion, jerk, and drift. Tremulous motion is superimposed onto the non-tremulous motion [Riley 87].

Compensation for the tremulous motion of the tool tip is achieved by manipulating the tool by the manipulator. The information that the manipulator requires for the compensation for tremulous motion of the tip is the tremulous motion of the tip with respect to non-tremulous motion of the tip. Base on this fact, an algorithm is developed to obtain tremulous motion by filtering out the non-tremulous motion using filters. Therefore, the motion that the algorithm provides is the tremulous motion with respect to the non-tremulous motion. To describe the motion obtained from the algorithm, a frame is chosen in such a way that the origin of the frame is attached to the non-tremulous motion of the tip. The orientation of the frame is defined the same as that of the body frame. The frame is named as “steady tip frame”, $\{S\}$. In another perspective, the frame $\{S\}$ can be considered a fixed frame since the motion information of the frame is filtered out by the algorithm.

The description of the tremulous motion in the following sections will be with respect to this frame $\{S\}$. The locations of the frame $\{S\}$, and the body frame $\{B\}$ which is attached to the instrument body are illustrated in Fig. 4.2. The displacement of the tip that the manipulator is to move for the compensation, ${}^M P = \begin{bmatrix} {}^M P_x & {}^M P_y & {}^M P_z \end{bmatrix}^T$ is negative of the position of the tip due to tremor with respect to $\{S\}$, which can be known by the algorithm which is described in the following session.

$${}^M P = -{}^S P_e \quad (4.1)$$

Where ${}^S P_e = \begin{bmatrix} {}^S P_{e_x} & {}^S P_{e_y} & {}^S P_{e_z} \end{bmatrix}$ is the vector position of the tip $\{e\}$ due to tremor with respect to $\{S\}$.

Fig. 4.2. Illustration of the locations of the frames $\{S\}$, and $\{B\}$ on the instrument which is shaking due to tremor. The symbol $\{e\}$ represents location of the tool tip.

4.2.2. Obtaining tremor acceleration at the tip

The block diagram of the algorithm to obtain tremulous acceleration of the tip with respect to the steady tip frame is shown in Fig. 4.3. The steps involved in the algorithm are as follows.

Step 1: Acceleration readings from the accelerometers are filtered using low-pass filters (LPF).

Step 2: The outputs of the LPFs are subtracted from their respective inputs.

Step 3: The results of the subtraction are filtered using Weighted Fourier Linear Combiner (WFLC)* [Riviere 98] or Bandlimited Multiple Fourier Linear Combiner (BMFLC)* [Veluvolu 07] which are essentially zero-phase band-pass filters to obtain tremulous accelerations at the accelerometer locations.

Step 4: Tremulous acceleration of the tip with respect to $\{S\}$ is obtained by rigid body kinematic computation using information on tremulous accelerations at accelerometer locations and angular acceleration information. Calculations of angular acceleration from accelerations at accelerometer locations are described in Chapter 5 and Chapter 6.

* The block diagrams and operations of WFLC and BMFLC are presented in Chapter 8.

Fig. 4.3. Block diagram of the algorithm to obtain tremulous acceleration of the instrument tool tip.

* The block diagrams and operations of WFLC and BMFLC are presented in Chapter 8.

The need for filtering (Step 1, 2, and 3)

The acceleration reading from an accelerometer results from two acceleration components: motion-induced acceleration and gravity. Motion-induced acceleration consists of accelerations due to intended motion, physiological tremor, low-frequency drift, and jerk. Since only tremor acceleration is of interest, other components including gravity have to be removed from the acceleration reading. A possible solution is to employ a linear band-pass filter such as a Butterworth filter having a pass-band covering the tremor band of 8-12 Hz. However, a linear band-pass filter changes the phase characteristics of the input signal, which is undesirable in this application of real-time compensation. Hence, zero-phase band-pass filters such as WFLC or BMFLC has to be employed.

A WFLC or BMFLC filter cannot produce accurate outputs if the input signal contains large magnitude frequency components outside the pass-band comparing to the magnitude of components within the pass-band (tremor magnitude). Therefore, to

reduce the magnitude of the low-frequency components without affecting the phase characteristics of the tremor signal, subtraction of LPF output from the acceleration reading is carried out before the use of WFLC or BMFLC filter as shown in Fig. 4.3.

The use of an LPF is to obtain the low frequency component's magnitude so that it can be removed from the acceleration reading. The filter output is subtracted from the filter input (acceleration reading from the accelerometer) to remove low frequency components such as gravity, intended motion, and drift. The resultant signal after subtraction contains only physiological tremor, accelerometer noise, and residual low-frequency acceleration, which is shown in Fig. 4.4(a). The residual low-frequency acceleration results from the phase-lag introduced by the LPF.

One might think that the subtraction of LPF output from the acceleration reading is the same as the use of a high-pass filter (HPF). However, the use of a high-pass filter causes the output of the filter to have phase-lead while the phase of the output signal obtained from the subtraction is almost the same as the original tremor signal as shown in Fig. 4.4(b). In the figure, the symbol ϕ shows the phase difference between the tremor and the HPF output. The offset α is due to the residual low-frequency acceleration. The signals shown in the figure are simulated ones. Filter order used for both LPF and HPF filters is four. The cut-off frequencies used for the HPF and LPF are 6 Hz and 3 Hz respectively.

(a) A simulation plot showing tremor signal, low-pass filter (LPF) input, and LPF output. LPF input is acceleration reading consisting of tremor signal and low-frequency component

(b) Simulation plot of result of subtraction of LPF output from LPF input, and that of HPF output.

Fig. 4.4. Simulation plots showing the phase of the signal resulting from subtraction of LPF output from acceleration reading (LPF input – LPF output) has negligible difference from that of the original signal (tremor) while that of the signal at HPF output has significant difference (ϕ).

The use of the low-pass filter introduces a phase-lag in the output signal meaning that there is an error in the output. The amount of error due to the phase-lag depends on rate of change in instrument orientation due to intention. It is safe to assume that the rate of change in instrument orientation due to intended motion is slow in

micromanipulation tasks and hence the amount of error is negligible. Dashed line in the plot of Fig. 4.4(a) shows the low-frequency component (gravity) whose magnitude is increasing due to changes in instrument orientation (accelerometer orientation) at the rate of about 3 deg/s. Dotted line in the same plot shows the error in LPF output or residual low-frequency acceleration due to the phase-lag. Due to the presence of residual low-frequency acceleration, the resultant signal from the subtraction has some offset α in magnitude comparing to the tremor as shown in dashed line in Fig. 4.4(b). However, the phase of the resultant signal is almost the same as that of the original tremor signal and the amount of offset as a result of the error due to the phase-lag has negligible effect on the estimation of tremor motion using a BMFLC filter as shown in Fig. 4.5.

The BMFLC filter removes accelerometer noises, and residual low-frequency component which exists after the subtraction due to the phase lag introduced by the low-pass filter. The output of the BMFLC filter is therefore the tremulous acceleration at the accelerometer location. If the acceleration reading is used as an input to the BMFLC filter without removing the low-frequency component from the acceleration reading, the filter cannot perform well. This is shown in dotted line in Fig. 4.5.

Fig. 4.5. Simulation plot showing that BMFLC filter can produce the output similar to the tremor signal when its input is LPF input – LPF output. The offset α in figure 4.4 has been removed. The plot also shows that BMFLC output is different from the tremor (has some offset β) if its input is the acceleration reading (i.e., including low-frequency component).

Kinematics (Step 4)

Once the tremulous accelerations at all accelerometer locations are known, tremulous acceleration of the tip is calculated by kinematic computation. The kinematic computation employed is based on the following principle.

If accelerometers are fixed on a rigid body, the total accelerations, $A_i = [a_{i_x} \ a_{i_y} \ a_{i_z}]^T$, each accelerometer at $\{i\}$ in rigid body frame $\{B\}$ senses include the inertial acceleration of the body, A_{IN} , the gravity, G , and the rotation-induced accelerations: the centripetal acceleration, $A_{i/C}$, and the tangential acceleration, $A_{i/T}$,

$$A_i = A_{IN} + G + A_{i/C} + A_{i/T}, i = 1, 2, \dots, n. \quad (4.2)$$

$$A_i = A_{IN} + G + \underbrace{\Omega \times (\Omega \times R) + \alpha \times R}_{\text{Rotation induced accelerations}} \quad (4.3)$$

where all the variables are relative to the body frame $\{B\}$, $\Omega = [\omega_x \ \omega_y \ \omega_z]^T$ is the angular velocity vector with respect to body frame, $\{B\}$, R is the vector from the unknown instantaneous center of rotation to the accelerometer location $\{i\}$, n is the number of accelerometers.

Taking the difference between the acceleration readings at two different locations, for simplicity and without losing generality, accelerometer locations at $\{1\}$ and $\{2\}$ are taken, the non-rotation induced acceleration components A_{IN} and G are eliminated since the linear inertial acceleration of the body and the gravity should be identical at different locations, and rotation-induced accelerations are obtained.

$$A_{12} = A_2 - A_1 = \Omega \times (\Omega \times P_{12}) + \alpha \times P_{12} \quad (4.4)$$

where $P_{12} = [p_{12_x} \ p_{12_y} \ p_{12_z}]^T$ is position vector from $\{1\}$ to $\{2\}$ with respect to $\{B\}$. $\alpha = [\alpha_x \ \alpha_y \ \alpha_z]^T$ is angular acceleration vector with respect to $\{B\}$.

Rearranging (3),

$$A_2 = A_1 + A_{12} = A_1 + \Omega \times (\Omega \times P_{12}) + \alpha \times P_{12} \quad (4.5)$$

According to (4.5), acceleration at location $\{2\}$ can be known if acceleration reading at $\{1\}$, angular velocity, and angular acceleration are known. Therefore, in

general, calculation of motion-induced acceleration at a point $\{p\}$, $A_p = [a_{p_x} \ a_{p_y} \ a_{p_z}]^T$, which is some known distance away from an accelerometer at location $\{i\}$ is as follows.

$$A_p = A_i + A_{ip} = A_i + \Omega \times (\Omega \times P_{ip}) + \alpha \times P_{ip} \quad (4.6)$$

where $P_{ip} = [p_{ip_x} \ p_{ip_y} \ p_{ip_z}]^T$ is the displacement vector from locations $\{i\}$ and $\{p\}$, $A_{ip} = A_p - A_i$.

As will be discussed in section 5.6, the angular velocities of the instrument are found to be very small in micromanipulation tasks [Latt 08a], and hence the centripetal acceleration terms $\Omega \times (\Omega \times P_{ip})$ in (4.6) can be ignored, and (4.6) becomes

$$A_p = A_i + A_{ip} = A_i + \alpha \times P_{ip} \quad (4.7)$$

Expanding (4.7) into components form,

$$a_{p_x} = a_{i_x} + p_{ip_z} \alpha_y - p_{ip_y} \alpha_z \quad (4.8.a)$$

$$a_{p_y} = a_{i_y} + p_{ip_x} \alpha_z - p_{ip_z} \alpha_x \quad (4.8.b)$$

$$a_{p_z} = a_{i_z} + p_{ip_y} \alpha_x - p_{ip_x} \alpha_y \quad (4.8.c)$$

Likewise, the tremulous acceleration of the tip with respect to $\{S\}$, ${}^S A_e = [{}^S a_{e_x} \ {}^S a_{e_y} \ {}^S a_{e_z}]^T$ is calculated using tremulous accelerations at accelerometer locations as shown in (4.9.a-4.9.c). Placement of accelerometers in Micron is shown in Fig. 4.6.

$${}^S a_{e_x} = {}^S a_{3_x} + {}^B p_{3e_z} {}^B \alpha_y - {}^B p_{3e_y} {}^B \alpha_z \quad (4.9.a)$$

$${}^S a_{e_y} = {}^S a_{3_y} + {}^B p_{3e_x} {}^B \alpha_z - {}^B p_{3e_z} {}^B \alpha_x \quad (4.9.b)$$

$${}^S a_{e_z} = {}^S a_{1_z} + {}^B p_{1e_y} {}^B \alpha_x - {}^B p_{1e_x} {}^B \alpha_y \quad (4.9.c)$$

where ${}^S a_{i_d}$, $i = 1, 2$, or 3 , and $d = x, y$, or z , is tremor acceleration at accelerometer location $\{i\}$ in d axis. ${}^B p_{ie_d}$, $i = 1, 2$, or 3 , and $d = x, y$, or z , is the distance from location $\{i\}$ to the tip along d axis. Since ${}^B p_{3e_x} = {}^B p_{3e_y} = 0$, (4.9.a-4.9.c) can be simplified as:

$${}^S a_{e_x} = {}^S a_{3_x} + {}^B p_{3e_z} {}^B \alpha_y \quad (4.10.a)$$

$${}^S a_{e_y} = {}^S a_{3_y} - {}^B p_{3e_z} {}^B \alpha_x \quad (4.10.b)$$

$${}^S A_{e_z} = {}^S a_{1_z} + {}^B p_{1e_y} {}^B \alpha_x - {}^B p_{1e_x} {}^B \alpha_y \quad (4.10.c)$$

Fig. 4.6. A schematic diagram of position and orientation of accelerometers in the instrument.

Once the tremulous acceleration of the tip is known, the tremulous position of the tip (the position of the tip due to tremor) ${}^S P_e = \begin{bmatrix} {}^S P_{e_x} & {}^S P_{e_y} & {}^S P_{e_z} \end{bmatrix}^T$ is obtained by analytical double-integration of the tremulous acceleration of the tip. Analytical integration will be presented in 8.7. The analytical integration can be used due to rhythmic nature of the tremor [Elble 90] and the merit of BMFLC [Veluvolu 07] or WFLC filters [Riviere 98].

4.3. Experimental Setup

To prove that the algorithm works, a prototype of sensing sub-system of Micron, which has been developed to gain better understanding on sensing performance of

Micron, is mounted on a nanopositioning system (P-561.3CD, Physik Instrumente, Germany) as shown in Fig. 4.7(a). Translational sinusoidal motion (at 10Hz, 50 μm amplitude) of the instrument is produced by nanopositioning system while the system is being rotated manually as shown in Fig. 4.7(b).

Translational motion is to simulate the tremor and rotation is to make the instrument orientation changes to prove that the algorithm works and can estimate the tremor well for different instrument orientations. The zero-phase band-pass filter employed in the experiment is BMFLC. The results of the estimation error are shown in Table 4.1.

Fig. 4.7. (a) Picture of Micron instrument on nanopositioning system, (b) schematic diagram of how the system is rotated.

Acceleration output from one of the accelerometers inside the sensing sub-system is acquired at 1000Hz sampling rate. Fig. 4.8 (a) shows the acceleration reading from the accelerometer when the system is rotated at a rate of about 3 deg/s, (b) shows acceleration output of a BMFLC filter when the input is the acceleration reading shown in (a), and (c) shows the position output from analytical double-integration using acceleration from (b). Fig. 4.9 shows the position error obtained by comparing the

ground truth signal known from the nanopositioning system and the position output in Fig. 4.8(c).

Fig. 4.8. Performance of a BMFLC filter: (a) Acceleration reading from accelerometer (used as input to the BMFLC filter), (b) Output of the BMFLC filter (Tremor acceleration) when the input is acceleration shown in (a), and (c) Tremor position after analytical double-integration of (b).

Fig. 4.9. Error in tremor position estimation when input to the BMFLC filter is without the removal of low-frequency acceleration component.

Fig. 4.10 shows the outputs of an LPF and a zero-phase LPF. The zero-phase LPF which is implemented using Matlab routine “FILTfilt” is used as a ground-truth in the calculation of the error in the LPF output. Fig. 4.11 shows error in low-frequency acceleration component estimation due to phase lag of fourth-order low-pass filter with cutoff frequency of 3 Hz for the change in orientation at the rate of about 3 deg/s. Fig. 4.12 (a) shows resultant acceleration after LPF output is subtracted from the acceleration reading, (b) and (c) respectively show (tremor) acceleration and position outputs. The position error is shown in Fig. 4.13.

Fig. 4.10. Outputs of an LPF and a zero-phase LPF.

Fig. 4.11. Error in low-frequency acceleration estimation due to phase lag of fourth-order low pass filter with cutoff frequency of 3 Hz for the change in orientation of 3 deg/s.

Fig. 4.12. Performance of a BMFLC filter : (a) Acceleration (used as input to the BMFLC filter) after subtraction of LPF output from acceleration reading, (b) Output of the BMFLC filter (Tremor acceleration) when the input is acceleration shown in (a), and (c) Tremor position after analytical double-integration of (b).

Fig. 4.13. Error in tremor position estimation when input to the BMFLC filter is without the low-frequency acceleration component.

4.4. Results and Discussion

The effect of the phase lag due to a low-pass filter on tremor estimation is negligible since it is only 0.28% and 1.98% for root mean square error (RMSE) and maximum error respectively as shown in Table 4.1. The percentage error is calculated using 35.36 μm which is the root mean square value of the sinusoidal motion with amplitude of 50 μm generated by nanopositioning system.

Comparing the errors shown in Fig. 4.9 and Fig. 4.13, the error in Fig. 4.9 is larger than that in Fig. 4.13 because of the inaccurate output of the BMFLC filter due to the presence of a large magnitude low-frequency component in the input to the filter. The error decreases significantly when the large magnitude low-frequency component is removed using a low-pass filter in the algorithm.

The error that exists even after the removal of low-frequency component is probably due to the performance of BMFLC filter. By properly tuning the BMFLC filter, the error can further be reduced.

Table 4.1 Error in tremor position estimation with and without the removal of low-frequency acceleration component

	I	II	III	IV
RMSE (μm)	5.1	2.4	2.3	0.1
RMSE (%)	14.42	6.79	6.51	0.28
Max Error (μm)	21.2	8.9	8.2	0.7
Max Error (%)	59.96	25.17	23.19	1.98

I - Without the removal of low-frequency component

II - With the removal of the low-frequency component using a normal LPF

III - With the removal of the low-frequency component using a zero-phase offline LPF

IV - Error due to phase lag

4.5. Conclusion

It has been shown that the tip motion with respect to the world frame is not necessary to compensate for the tremulous motion of the tip. A frame named “steady tip

frame” has been introduced for ease of representation of tremulous motion of the tip with respect to the non-tremulous motion of the tip. An algorithm to sense physiological tremor at the instrument tip with respect to the steady tip frame without having to know complete orientation of the instrument (i.e., with respect to the world frame) has been proposed and tested with real data. The preliminary evaluation and results raise chances of success in 3 DOF sensing of instrument tip tremulous motion using only accelerometers.

The steps involved in the algorithm described in this chapter serve as fundamental steps details of which are presented in chapter 9. The advantages obtained due to the algorithm which the author proposed is summarized as follows. Since a magnetometer is not required, on-site magnetometer calibration, which is prone to errors in case of the environment where magnetic field is changing or there are magnetic interference, is no longer necessary. The instrument can become light in weight, compact in size, and reliable in terms of lesser components due to the absence of the magnetometer.

4.6. References

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Chapter 5 : Placement of Accelerometers for Highest Possible Angular Sensing Resolution

This chapter begins with description of research efforts on detection of angular motion of a rigid body using only accelerometers. It highlights lack of literature or research on effect of accelerometer placement on motion sensing resolution. Moreover, it points out Micron's inadequate sensing resolution due to the non-optimal placement of accelerometers. It then proposes propositions for the placement of accelerometers to yield highest possible sensing resolution in micromanipulation. The propositions are proved by mathematics. Based on the propositions, two designs of accelerometer placement in the hand-held instrument are proposed. The improvement in sensing resolution (by reduction of sensing noise) in the proposed designs comparing to that in Micron placement design is confirmed with experimental data. Kinematics formulae employed in proving the propositions and kinematics described in chapter 4 neglect the centripetal acceleration by assuming that angular velocities in micromanipulation tasks are very small. The validity of the assumption is confirmed by conducting experiments. At the end of the chapter, contributions that the author has made are summarized.

5.1. Introduction

The concept of the viability of the use of linear accelerometers to measure angular motion of a rigid body is introduced in 1965 [DiNapoli 65]. A number of researchers contribute towards the improvement of the original concept [Schuler 67; Mital 79]. There seem to be little progress in this area of rigid body motion detection using accelerometers from 1979 until 1994 when an original cube configuration of placement of six accelerometers, the minimum number of accelerometers required to detect six degree-of-freedom (DOF) motion of a rigid body [Padgaonkar 75], is introduced by Chen *et al.* [Chen 94]. Several researchers have been working on the improvement of gyro-scope free inertial navigation systems (GF-INS) based on the original cube configuration [Wang 03; Tan 05; Akeila 08; Kao 08].

The position and velocity estimates of an inertial navigation system (INS) are obtained by integration of the acceleration data. All of the INSs including GF-INSs suffer from position drift owing to numerical integration, and the presence of inherent

errors at the output of the sensors such as bias and noise. A number of researchers have been proposing methods to estimate the bias accurately. For example, a model is proposed by Ang *et al.* [Ang 07] to eliminate cross-axis effect of an accelerometer and estimate its bias so that the drift caused by the bias is as little as possible. The drift or inaccuracy of pose estimation is normally corrected by external aids such as Global Positioning System (GPS) in INSs.

While the bias or zero-offset of an accelerometer or a gyroscope causes position or orientation drift which affects accuracy of pose estimation, the sensor noise not only affects the accuracy of the estimation but it also affects the precision of the estimation. In some applications such as sensing vibration of an object and sensing of tremulous motion of an instrument tool tip, precision of motion sensing is vital and drift which degrades accuracy of absolute pose estimation is not a concern. The reason is that high-frequency motion is of interest in these applications and drift which is a low-frequency component is not of interest. As far as the author's knowledge is concerned, there is little or no literature or research on the effect of accelerometer placement on motion sensing resolution/precision.

The use of the cube configuration is neither feasible nor yields highest possible sensing resolution in some applications owing to space constraints: the space available is not in cube shape. In these situations, researchers have to seek for other accelerometer placements within the given space constraints [Onodera 08; Ang 04]. The placements might not be optimal in terms of accuracy and resolution. There are no so-called golden rules from the view point of precision for the general placement of accelerometers to detect six DOF motion of a rigid body. For a given same space available, three different accelerometer placement configurations involving six accelerometers to detect six DOF motion are shown in Fig. 5.1. All the three placement configurations can provide 6 DOF motion sensing. The configuration on the right is the cube configuration introduced by Chen *et al.* [Chen 94]. However, it is not obvious which one would provide the highest angular sensing resolution.

Fig. 5.1. Three different configurations of accelerometer placement within the same space available to detect six DOF motion.

It is stated in [Riviere 03] that sensing resolution of Micron is poor. This might have resulted from non-optimal placement of accelerometers in the instrument. Therefore, it is necessary to find out and propose placements of accelerometers to yield high sensing resolution without having to restrict the design to the cube configuration, thereby allowing the highest possible sensing resolution for a given space constraint.

5.2. Propositions for Placements of Accelerometers

The propositions for the placement of the accelerometers to have the highest possible angular acceleration resolution (or the lowest possible angular acceleration sensing noise) in micromanipulation tasks are as follows.

Proposition (i): Separation distance along a principal axis between the two sensing axes of the accelerometers having the same sensing axis direction must be as large as possible. (d_1 in Fig. 5.2 represents the separation distance)

Proposition (ii): The two accelerometers having the same sensing axis direction must be placed in a way that their sensing axes form a plane perpendicular to the principal axis about which the angular acceleration is calculated. The placement that does not satisfy (ii) will require more accelerometer outputs in the calculation of a particular angular acceleration resulting in more noise in the angular acceleration.

Fig. 5.2. Illustration of the separation distance between the two sensing axes.

5.2.1. The proof of proposition (i)

For simplicity and without losing generality, the accelerometers placement in Fig. 5.2 is assumed. The total accelerations, A_i , each accelerometer at $\{i\}$ senses include the inertial acceleration of the body, A_{IN} , the gravity, G , and the rotation-induced accelerations: the centripetal acceleration, $A_{i/C}$, and the tangential acceleration, $A_{i/T}$,

$$A_i = A_{IN} + G + A_{i/C} + A_{i/T}, i=1,2; \quad (5.1a)$$

$$A_i = A_{IN} + G + \underbrace{\Omega \times (\Omega \times R) + \alpha \times R}_{\text{Rotation-induced accelerations}} \quad (5.1b)$$

where all the variables are relative to the body frame $\{B\}$, $\Omega = [\omega_x \ \omega_y \ \omega_z]^T$ is the angular velocity vector, R is the vector from the unknown instantaneous center of rotation to the point of sensing, and $\alpha = [\alpha_x \ \alpha_y \ \alpha_z]^T$ is angular acceleration vector. The symbol $\times$ represents a cross-product operation.

Taking the difference between the accelerations readings at $\{1\}$ and $\{2\}$, the non-rotation induced acceleration components A_{IN} and G are eliminated, since the linear inertial acceleration of the body and the gravity should be identical at different locations,

$$A_{12} = A_2 - A_1 = \Omega \times (\Omega \times P_{12}) + \alpha \times P_{12} \quad (5.2)$$

where $P_{12} = [p_{12_x} \ p_{12_y} \ p_{12_z}]^T$, is position vector from $\{1\}$ to $\{2\}$. In micromanipulation tasks, the angular velocity of the instrument is almost always very small [Latt 08]. Thus the Ω^2 centripetal acceleration terms in (5.2) may be ignored,

$$A_{12} = A_2 - A_1 = \alpha \times P_{12} = [\bullet \ \bullet \ a_{12_z}]^T \quad (5.3)$$

where the symbol $\bullet$ denotes undefined quantity. In component-form, (5.3) can be written as

$$\begin{bmatrix} \bullet \\ \bullet \\ a_{2_z} - a_{1_z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{12_x} \\ P_{12_y} \\ P_{12_z} \end{bmatrix} = \begin{bmatrix} 0 \\ \alpha_z P_{12_x} \\ -\alpha_y P_{12_x} \end{bmatrix} \quad (5.4)$$

$$\alpha_y = \frac{a_{2_z} - a_{1_z}}{-P_{12_x}} = \frac{a_{2_z} - a_{1_z}}{d_1} \quad (5.5)$$

From (5.5), standard deviation of noise of angular acceleration about Y_B axis, σ_{α_y} can be written as

$$\sigma_{\alpha_y} = \frac{\sqrt{\sigma_{2_z}^2 + \sigma_{1_z}^2}}{d_1} \quad (5.6)$$

where $\sigma_{2_z}^2$ and $\sigma_{1_z}^2$ are variances of noise outputs of accelerometers. The equation (5.6) is obtained from (5.5) using a random variables theorem. The random variables theorem used is stated as follows:

For $X_1, X_2, \dots, X_n$ be n mutually independent normal random variables with means $\mu_1, \mu_2, \dots, \mu_n$ and variances $\sigma_1^2, \sigma_2^2, \dots, \sigma_n^2$ respectively, if $Y = \sum_{i=1}^n C_i X_i$ then the mean of Y , μ_y , and variance of Y , σ_y^2 respectively are

$$\mu_y = \sum_{i=1}^n C_i \mu_i \text{ and } \sigma_y^2 = \sum_{i=1}^n C_i^2 \sigma_i^2 \quad (5.7)$$

From (5.6), the noises of a_{2_z} and a_{1_z} (accelerometer outputs noises) are fixed. Therefore, to minimize σ_{α_y} , separation distance (d_1) between accelerometer $\{1\}$ and $\{2\}$ must be maximized and hence the proposition (i) is proven.

5.2.2. The proof of proposition (ii)

In Fig. 5.3, sensing axes of accelerometers do not form a plane perpendicular to the principal axis about which the angular acceleration is calculated.

$$A_{12} = A_2 - A_1 = \alpha \times P_{12} \quad (5.8)$$

$$\begin{bmatrix} \bullet \\ \bullet \\ a_{2z} - a_{1z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{12x} \\ P_{12y} \\ P_{12z} \end{bmatrix} = \begin{bmatrix} \alpha_y P_{12z} - \alpha_z P_{12y} \\ -\alpha_x P_{12z} + \alpha_z P_{12x} \\ \alpha_x P_{12y} - \alpha_y P_{12x} \end{bmatrix} \quad (5.9)$$

$$\alpha_y P_{12x} = \alpha_x P_{12y} - (a_{2z} - a_{1z}) \quad (5.10)$$

$$\alpha_y = \frac{\alpha_x d_2}{d_1} + \frac{a_{2z} - a_{1z}}{d_1} \quad (5.11)$$

$$\sigma_{\alpha_y} = \frac{\sqrt{(d_2 \sigma_{\alpha_x})^2 + \sigma_{2z}^2 + \sigma_{1z}^2}}{d_1} \quad (5.12)$$

Fig. 5.3. Illustration showing the accelerometer placement not conforming to the proposition (ii). Accelerometer sensing axes do not form a plane perpendicular to Y_B axis.

Comparing equation (5.6) to (5.12), (5.12) contains one more term $\frac{\sqrt{(d_2 \sigma_{\alpha_x})^2}}{d_1}$ which causes extra noise in angular acceleration about Y_B axis (α_y).

The amount of extra noise depends on the ratio of d_1 and d_2 , and noise in α_x . If the two sensing axes of the accelerometers are on the plane, d_2 becomes zero and extra noise is eliminated, and hence proposition (ii) is proven.

In sections 5.3, 5.4, and 5.5, the importance of following the propositions to obtain highest possible angular sensing resolution is shown by comparing angular acceleration formulae and angular acceleration noise obtained from Micron accelerometer placement (the placement which does not following the proposition (ii)) with those obtained from accelerometer placements which follow the propositions.

5.3. Placement of Accelerometers in Micron and Angular Acceleration Derivation

There are 3 dual-axis accelerometers (6 accelerometers) in Micron. Placement of accelerometers in Micron is shown in Fig. 5.4.

(a) Isometric View

(b) View from the tip (or the distal end of the instrument)

Fig. 5.4. Placement of accelerometers in Micron with $d_1 = 20.2$ mm, $d_2 = 14.3$ mm, and $d_3 = 116$ mm.

The value of d_3 is about six times larger than that of d_1 . The placement does not comply with the proposition (ii) and hence the angular acceleration noise could be reduced further.

Taking the difference between the accelerations readings at {1}, {2} and {3}, and neglecting centripetal acceleration, the following equations are obtained.

$$A_{13} = A_3 - A_1 = \alpha \times P_{13} = \begin{bmatrix} a_{13_x} & \bullet & \bullet \end{bmatrix}^T \quad (5.13.a)$$

$$A_{23} = A_3 - A_2 = \alpha \times P_{23} = \begin{bmatrix} \bullet & a_{23_y} & \bullet \end{bmatrix}^T \quad (5.13.b)$$

$$A_{12} = A_2 - A_1 = \alpha \times P_{12} = \begin{bmatrix} \bullet & \bullet & a_{12_z} \end{bmatrix}^T \quad (5.13.c)$$

where $P_{ij} = \begin{bmatrix} p_{ij_x} & p_{ij_y} & p_{ij_z} \end{bmatrix}^T$, $i, j = 1, 2, 3$, is position vector from {i} to {j}, which are known values from the hardware design and calibrations. Equation (5.13.a) can be written in component-form as

$$\begin{bmatrix} a_{3_x} - a_{1_x} \\ \bullet \\ \bullet \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{13_x} \\ P_{13_y} \\ P_{13_z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} -(d_1 - d_2) \\ d_2 \\ d_3 \end{bmatrix} \quad (5.14)$$

$$a_{3_x} - a_{1_x} = \alpha_y d_3 - \alpha_z d_2 \quad (5.15)$$

Equation (5.13.b) can be written in component-form as

$$\begin{bmatrix} \bullet \\ a_{3_y} - a_{2_y} \\ \bullet \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{23_x} \\ P_{23_y} \\ P_{23_z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} d_2 \\ -(d_1 - d_2) \\ d_3 \end{bmatrix} \quad (5.16)$$

$$a_{3_y} - a_{2_y} = \alpha_z d_2 - \alpha_x d_3 \quad (5.17)$$

Equation (5.13.c) can be written in component-form as

$$\begin{bmatrix} \bullet \\ \bullet \\ a_{2_z} - a_{1_z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{12_x} \\ P_{12_y} \\ 0 \end{bmatrix} = \begin{bmatrix} -\alpha_z P_{12_y} \\ \alpha_z P_{12_x} \\ \alpha_x P_{12_y} - \alpha_y P_{12_x} \end{bmatrix} \quad (5.18)$$

$$a_{2_z} - a_{1_z} = \alpha_y d_1 + \alpha_x d_1 \quad (5.19)$$

Solving equation (5.15), (5.17), and (5.19),

$$\alpha_x = \frac{-(a_{3_x} - a_{1_x}) - (a_{3_y} - a_{2_y})}{2d_3} + \frac{a_{2_z} - a_{1_z}}{2d_1} \quad (5.20)$$

$$\alpha_y = \frac{(a_{3_x} - a_{1_x}) + (a_{3_y} - a_{2_y})}{2d_3} + \frac{a_{2_z} - a_{1_z}}{2d_1} \quad (5.21)$$

$$\alpha_z = \frac{(a_{2_y} - a_{1_y})}{2d_2} - \frac{d_3(a_{2_z} - a_{1_z})}{2d_2d_1} + \frac{(a_{3_x} - a_{1_x})}{2d_2} \quad (5.22)$$

As can be seen from (5.20-5.22), calculation of angular acceleration about each axis requires outputs of all the accelerometers resulting in more noise in the angular accelerations as a result of not satisfying proposition (ii). Noise will be reduced if the accelerometer placement satisfies the propositions. Therefore, new placement designs based on the propositions are proposed.

5.4. Proposed Designs

5.4.1. Proposed design I

The placement of accelerometers in the proposed design I is shown in Fig. 5.5. The values of d_1 and d_3 are the same as those in Micron design.

Angular accelerations about Y_B and Z_B axis are obtained from acceleration readings of accelerometers at location {1} and {2} as follows.

$$\alpha \times P_{12} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{12_x} \\ P_{12_y} \\ P_{12_z} \end{bmatrix} = \begin{bmatrix} \bullet \\ a_{2_y} - a_{1_y} \\ a_{2_z} - a_{1_z} \end{bmatrix}.$$

Since $P_{12_y} = P_{12_z} = 0$,

$$\alpha \times P_{12} = \begin{bmatrix} 0 \\ \alpha_z P_{12_x} \\ -\alpha_y P_{12_x} \end{bmatrix} = \begin{bmatrix} \bullet \\ a_{2_y} - a_{1_y} \\ a_{2_z} - a_{1_z} \end{bmatrix}$$

$$\alpha_z = \frac{a_{2_y} - a_{1_y}}{P_{12_x}} = \frac{a_{1_y} - a_{2_y}}{d_1} \quad (5.23)$$

$$\alpha_y = -\frac{a_{2_z} - a_{1_z}}{P_{12_x}} = \frac{a_{2_z} - a_{1_z}}{d_1} \quad (5.24)$$

Using α_z value from equation (5.23) and acceleration readings from accelerometers at location {2} and {3}, α_x is calculated as follows:

As in equation (3),

$$\alpha \times P_{23} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{23_x} \\ P_{23_y} \\ P_{23_z} \end{bmatrix} = \begin{bmatrix} \bullet \\ a_{3_y} - a_{2_y} \\ \bullet \end{bmatrix}$$

Since $P_{23_x} = \frac{d_1}{2}$, $P_{23_y} = 0$, $P_{23_z} = d_3$,

$$\alpha \times P_{23} = \begin{bmatrix} \alpha_y d_3 \\ -\alpha_x d_3 + \frac{\alpha_z d_1}{2} \\ -\alpha_y \frac{d_1}{2} \end{bmatrix} = \begin{bmatrix} \bullet \\ a_{3_y} - a_{2_y} \\ \bullet \end{bmatrix}$$

$$\alpha_x = \frac{a_{2_y} + \frac{\alpha_z d_1}{2} - a_{3_y}}{d_3} = \frac{a_{2_y} + \frac{a_{1_y} - a_{2_y}}{2} - a_{3_y}}{d_3}$$

$$\alpha_x = \frac{a_{1_y} + a_{2_y} - 2a_{3_y}}{2d_3} \quad (5.25)$$

Fig. 5.5. Proposed design I.

The noise in α_x and α_z are found to be lower than those present in Micron design. The noise in α_y is found to be higher than that in Micron design. The amount of noise present in each angular acceleration is known by using outputs of static accelerometers in the formulae. To improve α_y resolution, another design is proposed by adding one more single-axis accelerometer as shown in Fig. 5.6. Therefore, total number of accelerometers in the proposed design II is seven.

5.4.2. Proposed design II

Fig. 5.6. Proposed design II.

As in equation (5.3),

$$\alpha \times P_{34} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} P_{34_x} \\ P_{34_y} \\ P_{34_z} \end{bmatrix} = \begin{bmatrix} a_{4_x} - a_{3_x} \\ \bullet \\ \bullet \end{bmatrix}$$

since $P_{34_x} = P_{34_y} = 0, P_{34_z} = -d_3$

$$\alpha \times P_{34} = \begin{bmatrix} -\alpha_y d_3 \\ \alpha_x d_3 \\ 0 \end{bmatrix} = \begin{bmatrix} a_{4_x} - a_{3_x} \\ \bullet \\ \bullet \end{bmatrix}$$

$$\alpha_y = \frac{a_{3x} - a_{4x}}{d_3} \quad (5.26)$$

The equations for α_z and α_x are found to be the same as (5.23) and (5.25) respectively.

$$\alpha_z = \frac{a_{2y} - a_{1y}}{P_{12x}} = \frac{a_{1y} - a_{2y}}{d_1} \quad (5.27)$$

$$\alpha_x = \frac{a_{1y} + a_{2y} - 2a_{3y}}{2d_3} \quad (5.28)$$

5.5. Experiments and Results

To prove the propositions experimentally, a prototype of the sensing sub-system following the proposed design II accelerometer placement is developed. The prototype of the sensing sub-system is shown in Fig. 5.7. The complete instrument prototype with the sensing sub-system following the proposed design II accelerometer placement is called “MicronII”. Accelerometers employed are dual-axis miniature MEMS accelerometers (ADXL203, Analog Devices, USA).

Fig. 5.7. MicronII with the sensing sub-system according to design II placement of accelerometers.

Outputs of the accelerometers placed inside the instrument are acquired at 500 Hz sampling rate and are put in the equations (5.20-5.22), (5.23-5.25) and (5.26-5.28) to obtain angular accelerations for Micron design, proposed design I, and proposed design II respectively. Since the instrument is stationary, the ground truth angular acceleration is zero. Therefore, angular accelerations obtained from the equations are due to noise.

Angular acceleration noises obtained in different designs are shown in Fig. 5.8, Fig. 5.9, and Fig. 5.10.

Fig. 5.8. Angular acceleration noise in Micron design.

Fig. 5.9. Angular acceleration noise in proposed design I.

Table 5.1 shows comparison of standard deviation of angular acceleration noise of all the accelerometer placement designs. Table 5.2 shows amount of angular acceleration noise reduction by proposed designs.

The calculation of amount of noise reduction comparing to the design of placement in Micron is based on the following formula.

$$\frac{N_{pd} - N_{NDid}}{N_{pd}} \times 100\%$$

where N_{pd} and N_{NDid} , $d = x, y, \& z, i = I, \& II$, are standard deviations of noise in angular acceleration about d axis in Micron design and proposed design i respectively.

Fig. 5.10. Angular acceleration noise in proposed design II.

Table 5.1. Comparison of standard deviation (STD) of angular acceleration noise of all the accelerometer placement designs

STD of Angular Acceleration Noise	Micron Design	Proposed Design I	Proposed Design II
α_x (rad/s ²)	0.36	0.11	0.11
α_y (rad/s ²)	0.36	0.70	0.12
α_z (rad/s ²)	2.93	0.71	0.71
Vector α (rad/s ²)	2.97	1.00	0.73

Table 5.2. Angular acceleration noise reduction by proposed designs

Noise reduction	by Proposed Design I	by Proposed Design II
in α_x (%)	69.44	69.44
in α_y (%)	-94.44	66.67
in α_z (%)	75.77	75.77
in vector α (%)	66.33	75.42

To confirm that the angular acceleration formulae described in (5.20-5.28) are correct, an experiment is conducted. MicronII is placed on a one-degree of freedom rotating arm as shown in Fig. 5.11. The distal end of the sensing part is attached with a non-reflective black color rod. The tip of the rod is attached with a reflective ball whose location is sensed using M^2S^2 covered with black-color painted plates as shown in Fig. 5.12. M^2S^2 is rotated 90° to suit the experiment. The location of the ball with respect to the center of rotation of the rotating arm is known.

(a)

Fig. 5.11. A photo of (a) the instrument attached to the rotating arm, and (b) close-up view of the instrument tip placed near to the workspace of M^2S^2 .

Fig. 5.12. M^2S^2 covered with black-color painted plates and before it is rotated 90° to suit the experiment.

Rotational motion of the instrument at the frequency of about 3-4 Hz is produced by shaking the rotating arm horizontally back-and-forth. The rotational acceleration of the instrument is calculated from the M^2S^2 measured positions of the ball which is attached to the instrument. The measured position of the ball is double-differentiated to obtain the ball acceleration which is further divided by the distance from the location of the ball to the center of rotation to obtain angular acceleration. Since the angular acceleration obtained contains noise due to the differentiation, offline zero-phase band-pass filtering is performed using a 4th order band-pass filter having a pass-band of 1-5 Hz. The filtered angular acceleration is used as ground truth angular acceleration.

Plots of comparison of ground truth angular acceleration about Y_B axis and the calculated angular acceleration about Y_B axis obtained using the equation (5.26) and the acceleration readings from the accelerometers in the instrument are shown in Fig. 5.13. The calculated angular acceleration from the instrument is also filtered for evaluation using the same band-pass filter used in obtaining the ground truth angular acceleration. Experiment results for the angular sensing evaluation are shown in Table 5.3. Sixteen seconds length of data is used for the evaluation.

(a)

(b)

Fig. 5.13. Plots of comparison of ground truth angular acceleration about Y_B axis and calculated angular acceleration about Y_B axis using acceleration outputs from accelerometers inside the instrument: (a) without, and (b) with band-pass filtering the calculated angular acceleration.

RMS (rad/s²) is root mean squared value of ground truth angular accelerations. RMSE (rad/s²) are obtained by subtracting the ground-truth angular accelerations from the filtered calculated angular accelerations from MicronII. Sensing accuracies of MicronII are obtained as follows:

$$\text{Accuracy (\%)} = \frac{(\text{RMS}(\mu\text{m}) - \text{RMSE}(\mu\text{m}))}{\text{RMS}(\mu\text{m})} \times 100$$

Table 5.3 Experiment results for the angular sensing evaluation

	X axis	Y axis	Z axis
RMS (rad/s ²)	0.55	0.63	0.73
RMSE (rad/s ²)	0.04	0.05	0.13
Accuracy (%)	93.30	92.04	82.20

5.6. Study of the Instrument’s Angular Motion

It has been assumed in previous sections and in [Ang 03; Latt 09] that angular velocity in micromanipulation tasks is always very small and hence the on-line calculation of angular acceleration required in kinematics to get instrument tip motion is carried out analytically by neglecting centripetal acceleration. However, there is no quantitative experimental proof on the assumption and there is a lack of quantitative information on angular motion of the instrument due to physiological tremor during micromanipulation tasks. Centripetal acceleration at a point on a rigid body is directly proportional to the square of the angular velocity of the rigid body. Therefore, the assumption is clarified by calculating the angular velocities of instrument held by subjects performing representative micromanipulation tasks such as stationary and tracking tasks.

Apart from the clarification of the assumption, the knowledge on angular motion of the instrument due to physiological tremor in micromanipulation tasks can be useful in choosing sensors in hand-held tremor compensation instruments. For example, particular kinds of sensors such as gyroscopes or accelerometers can be used for angular sensing if it has high enough resolution to detect an angular motion of the instrument

due to tremor. For the above reasons, angular velocity and angular acceleration of the instrument due to tremor is measured.

5.6.1. Six DOF sensing system for the study

Sensing sub-system of MicronII is used for the study. A photo of the instrument being held by a subject performing a stationary test is shown in Fig. 5.14. The sampling rate of the sensing system is 1 kHz. The acceleration data sampled from accelerometers are processed to get angular velocity and angular acceleration and are filtered offline using a zero-phase band-pass filter having a pass-band range of 6-14 Hz to remove noise and other components rather than physiological tremor. The weight of the instrument is 60 gm. The diameter of the handle of the instrument is 20 mm.

Fig. 5.14. A subject performing a stationary test by holding the instrument at grip location B.

5.6.2. Numerical solution to get angular velocity

As described in section 5.2, taking the difference between the acceleration readings at {1} and {2}, {2} and {3}, and {3} and {4}, non-rotation induced acceleration are eliminated and the following equations are obtained.

$$A_{12} = A_2 - A_1 = \Omega \times (\Omega \times P_{12}) + \alpha \times P_{12} \quad (5.29)$$

$$A_{23} = A_3 - A_2 = \Omega \times (\Omega \times P_{23}) + \alpha \times P_{23} \quad (5.30)$$

$$A_{34} = A_4 - A_3 = \Omega \times (\Omega \times P_{34}) + \alpha \times P_{34} \quad (5.31)$$

where $P_{ij} = [p_{ijx} \ p_{ijy} \ p_{ijz}]^T$, $i=1,2$, or 3 ; $j=2,3$, or 4 , is position vector from $\{i\}$ to $\{j\}$, which are known values from the hardware design. $A_{ij} = [a_{ijx} \ a_{ijy} \ a_{ijz}]^T$ is the difference in acceleration readings between accelerometers at $\{i\}$ and $\{j\}$. Rewriting (6.2-6.4) into component form,

$$a_{12y} = p_{12x}\omega_x\omega_y - p_{12y}(\omega_x^2 + \omega_z^2) + p_{12z}\omega_y\omega_z + p_{12x}\frac{\omega_z - \omega_z(t-T_s)}{T_s} - p_{12z}\frac{\omega_x - \omega_x(t-T_s)}{T_s} \quad (5.32)$$

$$a_{12z} = p_{12x}\omega_x\omega_z + p_{12y}\omega_y\omega_z - p_{12z}(\omega_x^2 + \omega_y^2) - p_{12x}\frac{\omega_y - \omega_y(t-T_s)}{T_s} - p_{12y}\frac{\omega_x - \omega_x(t-T_s)}{T_s} \quad (5.33)$$

$$a_{23y} = p_{23x}\omega_x\omega_y - p_{23y}(\omega_x^2 + \omega_z^2) + p_{23z}\omega_y\omega_z + p_{23x}\frac{\omega_z - \omega_z(t-T_s)}{T_s} - p_{23z}\frac{\omega_x - \omega_x(t-T_s)}{T_s} \quad (5.34)$$

$$a_{34x} = -p_{34x}(\omega_y^2 + \omega_z^2) + p_{34y}\omega_x\omega_y + p_{34z}\omega_x\omega_z - p_{34y}\frac{\omega_z - \omega_z(t-T_s)}{T_s} - p_{34z}\frac{\omega_y - \omega_y(t-T_s)}{T_s} \quad (5.35)$$

Where t is the time index, and T_s is the sampling period. Time indexes for $a_{ijx}, a_{ijy}, a_{ijz}, \omega_x, \omega_y$, and ω_z are omitted for clarity of reading.

The angular velocity vector is found by solving the nonlinear system of equations (5.32-5.35) using trust-region dogleg algorithm in Matlab. Root mean square (RMS) noise of angular velocity about X and Y axes is (0.54 mrad/s) 0.03 deg/s and that about Z axis is (3.1 mrad/s) 0.17 deg/s.

5.6.3. Experiment

Ten normal subjects are asked to perform stationary and tracking tests which are described in this section. Three trials are carried out for each test per person. To see the effect of different gripping locations on angular motion of the instrument, subjects are asked to grip the instrument at two different locations; grip location “A” and grip location “B” as shown in Fig. 5.14 for both tests. The two locations, “A” and “B”, have separation distance of 25 mm.

Stationary

Subjects are asked to aim without touching at a black color spot on a white paper with the instrument tip while holding the instrument stationary for 20 s (see Fig. 5.14). Subjects' forearm and palm are fully supported during the trials.

Tracking

Subjects are asked to make the instrument tip follow a moving dot moving in a circular path having 5 mm diameter. A moving dot is created in software and shown on an LCD screen which is placed horizontally and facing upwards for the tracking tests. The speeds of the dot for slow tracking tests and fast tracking tests are 10 s/rev (0.25 mm/s) and 4 s/rev (0.625 mm/s) respectively; which are believed to be faster than the tool tip speed usually present in micromanipulation tasks such as microsurgeries and cell manipulations.

5.6.4. Results and discussion

Statistics of angular velocities of the instrument during the experiment are shown in Table 5.4-Table 5.15 while those of angular accelerations are shown in Table 5.16 and Table 5.17.

Table 5.4. *RMS* amplitudes of tremor angular velocity in stationary tests (Subjects held at grip location "A")

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0060	0.0016	0.0040	0.0098
Y	0.0076	0.0053	0.0034	0.0252
Z	0.0142	0.0066	0.0076	0.0369
Vector	0.0174	0.0082	0.0093	0.0457

Table 5.5. Maximum peak amplitudes of tremor angular velocity in stationary tests (Subjects held at grip location “A”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0224	0.0062	0.0133	0.0368
Y	0.0250	0.0157	0.0111	0.0765
Z	0.0511	0.0209	0.0292	0.1175
Vector	0.0579	0.0248	0.0343	0.1417

Table 5.6. *RMS* amplitudes of tremor angular velocity in stationary tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0063	0.0028	0.0028	0.0157
Y	0.0060	0.0024	0.0020	0.0131
Z	0.0126	0.0046	0.0058	0.0283
Vector	0.0153	0.0059	0.0067	0.0349

Table 5.7. Maximum peak amplitudes of tremor angular velocity in stationary tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0221	0.0096	0.0093	0.0584
Y	0.0218	0.0080	0.0102	0.0465
Z	0.0469	0.0173	0.0191	0.1001
Vector	0.0520	0.0189	0.0219	0.1100

Table 5.8. *RMS* amplitudes of tremor angular velocity in slow tracking tests (Subjects held at grip location “A”).

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0094	0.0037	0.0051	0.0166
Y	0.0088	0.0046	0.0033	0.0213
Z	0.0140	0.0056	0.0080	0.0299
Vector	0.0193	0.0074	0.0104	0.0380

Table 5.9. Maximum peak amplitudes of tremor angular velocity in slow tracking tests (Subjects held at grip location “A”).

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0366	0.0135	0.0191	0.0669
Y	0.0327	0.0150	0.0151	0.0816
Z	0.0541	0.0231	0.0287	0.1257
Vector	0.0626	0.0248	0.0338	0.1414

Table 5.10. *RMS* amplitudes of tremor angular velocity in slow tracking tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0099	0.0044	0.0044	0.0197
Y	0.0090	0.0059	0.0037	0.0260
Z	0.0129	0.0059	0.0066	0.0271
Vector	0.0186	0.0094	0.0088	0.0424

Table 5.11. Maximum peak amplitudes of tremor angular velocity in slow tracking tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0366	0.0140	0.0157	0.0651
Y	0.0338	0.0175	0.0123	0.0680
Z	0.0517	0.0269	0.0252	0.1311
Vector	0.0608	0.0281	0.0290	0.1362

Table 5.12. *RMS* amplitudes of tremor angular velocity in fast tracking tests (Subjects held at grip location “A”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0109	0.0048	0.0057	0.0212
Y	0.0104	0.0056	0.0056	0.0251
Z	0.0156	0.0075	0.0088	0.0363
Vector	0.0217	0.0105	0.0119	0.0490

Table 5.13. Maximum peak amplitudes of tremor angular velocity in fast tracking tests (Subjects held at grip location “A”).

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0424	0.0190	0.0184	0.1038
Y	0.0377	0.0170	0.0216	0.0869
Z	0.0570	0.0259	0.0326	0.1252
Vector	0.0664	0.0291	0.0355	0.1493

Table 5.14. *RMS* amplitudes of tremor angular velocity in fast tracking tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0098	0.0045	0.0051	0.0215
Y	0.0092	0.0036	0.0054	0.0167
Z	0.0132	0.0050	0.0075	0.0291
Vector	0.0188	0.0076	0.0106	0.0398

Table 5.15. Maximum peak amplitudes of tremor angular velocity in fast tracking tests (Subjects held at grip location “B”)

Direction	Average (rad/s)	STD (rad/s)	Min (rad/s)	Max (rad/s)
X	0.0362	0.0172	0.0195	0.0917
Y	0.0344	0.0126	0.0179	0.0570
Z	0.0503	0.0172	0.0289	0.1108
Vector	0.0595	0.0201	0.0332	0.1109

Table 5.16 *RMS* amplitudes of tremor angular acceleration in stationary tests (Subjects held at grip location “A”)

Direction	Average (rad/s ²)	STD (rad/s ²)	Min (rad/s ²)	Max (rad/s ²)
X	0.44	0.11	0.28	0.66
Y	0.58	0.29	0.29	1.54
Z	0.98	0.32	0.58	2.12
Vector	1.23	0.42	0.72	2.70

Table 5.17 Maximum peak amplitudes of tremor angular acceleration in stationary tests (Subjects held at grip location “A”)

Direction	Average (rad/s ²)	STD (rad/s ²)	Min (rad/s ²)	Max (rad/s ²)
X	1.82	0.59	1.04	3.32
Y	2.17	1.06	0.97	5.68
Z	3.98	1.33	2.17	7.50
Vector	4.56	1.62	2.43	9.58

There is no significant difference of tremor angular velocities/accelerations present in stationary and tracking tests. Tremor angular velocities/accelerations present when gripping at B are slightly lower than that present when gripping at A. Tremor angular velocities/accelerations about Z axis are found to be higher than those about X and Y axes. Mean of maximum peak amplitudes of tremor angular accelerations about Z axis presents in stationary test is about 4 rad/s² (see Table 5.17). The tangential acceleration of a point one unit away from the center of rotation is therefore $4 \times 1 = 4$ unit/s². Mean of maximum peak amplitudes of tremor angular velocities about Z axis is about 0.05 rad/s (please refer to Table 5.5, 5.7, 5.9, 5.11, 5.13, and 5.15). The centripetal acceleration of the same point is therefore $0.05 \times 0.05 \times 1 = 0.0025$ unit/s². Therefore, centripetal acceleration is only $0.0025/4 = 0.06\%$ of tangential acceleration and hence assumption of very small angular velocity to neglect centripetal acceleration in the analytical calculation of angular acceleration has been proved to be valid.

Fig. 5.15. *RMS* amplitude and maximum peak-amplitude of tremor angular velocity vectors.

5.7. Discussion

The RMSEs and accuracies in Table 5.3 show that the angular acceleration formulae (5.26-5.28) are correct although there are some errors. The errors might have resulted from the combined errors of the inaccurate positioning and orientation of the accelerometers, noises in the accelerometer outputs, accuracy and resolution of the measurement of the ball position by M^2S^2 . The positioning and orientation errors of the accelerometers can be known by using high precision rotation and translation stages and they can be calibrated. However, since the stages were not available to the authors when this research was carried out, these errors could not be identified and eliminated. That means the errors could have been reduced. The error in angular acceleration about Z_B

axis is more than those about X_B and Y_B axes. This might be due to more noise in angular acceleration about Z_B axis due to the smaller separation distance between the two accelerometers. Although the experiment is performed to show that the angular acceleration formulae for proposed design II are correct, those for proposed design I must also be correct since the derivation and assumption are the same.

In proposed design I, although noises in α_x and α_z have reduced by 69.44 percent and 75.77 percent respectively, noise in α_y is found have 94.44 percent more than the Micron design. This is due to the limited separation distance of d_1 in proposed design I (see equation (5.24)). Adding one more accelerometer as shown in Fig. 5.6 reduces noise in α_y by 66.67 percent.

Noise in α_z is the lowest possible since the separation distance d_1 is the maximum possible distance for the width of the instrument, and a_{2_y} and a_{1_y} form a plane perpendicular to the Z_B axis.

For α_y , a_{2_z} and a_{1_z} form a plane perpendicular to Y_B axis. For the same separation distance and the same orientations of the accelerometers, this placement yields the lowest noise in α_y . However, since the resolution is limited by the small separation distance d_1 , one more accelerometer is added at the origin of {B} in Proposed Design II. The new accelerometer a_{4_x} and existing a_{3_x} form the plane perpendicular to Y_B axis and the separation distance between them is d_3 which is much larger than d_1 .

Noise in α_x in proposed design I is not the lowest possible since sensing axes of a_{2_y} and a_{3_y} do not form a plane perpendicular to X_B axis. If the location of accelerometer at {2} were moved to the origin of {B}, then perpendicular plane would be formed and the noise in α_x would be the lowest possible. However, placing the accelerometer at the origin of {B} would increase the noise in α_z by about two times since separation distance between a_{1_y} and a_{2_y} to calculate α_z would become half.

Since extra noise due to not forming the perpendicular plane for α_x is small comparing to the amount of noise reduction in α_z , the placement can be considered optimal.

5.7.1. Redundancy and placement

One might think that extra noise reduction after the addition of seventh accelerometer is due to the redundancy. Although the noise reduction is due to redundancy, it is mainly due to the placement of accelerometers. It is explained below.

Generally, noise reduction by addition of more sensors is as follows:

$$\frac{1}{\sigma^2} = \frac{1}{\sigma_1^2} + \frac{1}{\sigma_2^2} + \dots + \frac{1}{\sigma_n^2} \text{, where } \sigma_i, i=1,2,\dots,n, \text{ is standard deviation of } i^{\text{th}}$$

sensor noise. σ is the standard deviation of the noise obtained due to redundancy.

Therefore, for the sensors having standard deviation of noise output of 10 mm/s², adding one more redundant sensor to existing 6 sensors would improve the resolution by 7.35 %. The calculation is shown below. The noise output of six redundant sensors is

$$\frac{1}{\sigma^2} = \frac{1}{\sigma_1^2} + \frac{1}{\sigma_2^2} + \dots + \frac{1}{\sigma_6^2}$$

$$\frac{1}{\sigma^2} = \frac{1}{10^2} + \frac{1}{10^2} + \dots + \frac{1}{10^2}$$

$\sigma = 4.08 \text{ mm/s}^2$. If one more redundant sensor is included, the noise will be

$$\frac{1}{\sigma^2} = \frac{1}{4.08^2} + \frac{1}{10^2}$$

$$\sigma = 3.78 \text{ mm/s}^2.$$

The noise reduction or resolution improvement is $\frac{(4.08-3.78)}{4.08} \times 100 = 7.35\%$.

However, due to the proper placement of one more accelerometer in proposed design II, the noise reduction or resolution improvement is $\frac{1-0.73}{1} \times 100 = 27\%$ (see the last row of Table 5.1). Therefore, the resolution improvement is mainly due to the placement and not due to the redundancy alone.

5.7.2. Noise in sensing of tool tip motion

It is reported that shaft lengths of microsurgical tools vary depending on the task and specialty. In general, a 50 to 100 mm shaft length is needed for ear surgery, and 100 to 160 mm for neurosurgery [Charles 96]. Accordingly, the displacement from the tool tip to the distal-end accelerometer (accelerometer {3} in Fig. 4.6 and Fig. 5.4) varies from 50 mm to 160 mm. To know the tip motion, multiplication of this displacement and angular acceleration is required in rigid-body kinematic computation as shown in (4.10.a-4.10.c).

Multiplication of a nominal displacement of 100 mm and the angular acceleration noise of 0.36 rad/s^2 present in Micron design will contribute the tip acceleration noise of about 37 mm/s^2 . The tip acceleration noise can be found using (4.10.a-4.10.c) as:

$$\sigma_{a_{e_x}} = \sqrt{\sigma_{a_{3_x}}^2 + ({}^B p_{3e_z} \sigma_{\alpha_y})^2} = \sqrt{10^2 + (100 \times 0.36)^2} = 37.36 \text{ mm/s}^2 \quad (5.36.a)$$

$$\sigma_{a_{e_y}} = \sqrt{\sigma_{a_{3_y}}^2 + ({}^B p_{3e_z} \sigma_{\alpha_x})^2} = \sqrt{10^2 + (100 \times 0.36)^2} = 37.36 \text{ mm/s}^2 \quad (5.36.b)$$

The equivalent amplitude of the tip acceleration with sinusoidal motion profile is therefore $\sqrt{2}\sigma_{a_{e_x}} = \sqrt{2}\sigma_{a_{e_y}} = \sqrt{2} \times 37.36$.

The noise values are quite significant since equivalent amplitude of tremor motion of 10 Hz in position domain is found to be about $13 \mu\text{m}$ as follows. Using (2.5) and equivalent amplitude of the tip acceleration of $\sqrt{2} \times 37.36$, amplitude of the tip motion due to tremor in position domain is

$$X = \frac{\sqrt{2} \times 37.36}{\omega^2} = 13 \mu\text{m} \quad (5.37)$$

The value does not meet the desired criterion of positioning accuracy of $\pm 10 \mu\text{m}$ [Charles 96]. On the other hand, using the angular acceleration noise in proposed designs, tip acceleration noise is found to be 14.14 mm/s^2 . The equivalent amplitude in position domain is

$$X = \frac{14.14\sqrt{2}}{\omega^2} = 5 \mu\text{m} \quad (5.38)$$

As can be seen from the above equation, the positioning accuracy criterion is now met by the proposed designs.

5.8. Conclusion

In this chapter, one of the contributions is propositions for the placement (location and orientation) of accelerometers in the instruments to be used in micromanipulation tasks to obtain the least possible angular acceleration noise. The placement of accelerometers should follow the propositions described to obtain lowest possible angular acceleration noise. The fact has been proven by proposing new designs of placement of accelerometers and comparing them with Micron accelerometer placement. Correctness of the angular acceleration formulae for the proposed designs has been shown by experiment. Significant reduction in noise level (i.e., significant improvement in sensing resolution) shows that the contribution is significant. The proposed propositions would serve as guidelines for future designs of placement of accelerometers within a given space constraint such as that of the slender instrument *ITrem* to obtain highest possible angular sensing resolution.

Another contribution to knowledge comes from the study of angular motion of a surgical instrument due to physiological tremor in micromanipulation tasks. Although immediate goal of the study is to see whether or not the assumption to neglect centripetal acceleration is a valid one, the knowledge obtained from the study can be helpful in future instrument designs.

5.9. References

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Chapter 6 : Compact Design of Placement of Accelerometers

This chapter describes a compact design of accelerometer placement which is based on the propositions proposed in the previous chapter. It presents the importance of the instrument being light in weight and compact in size for better manipulation. Following that, kinematics behind the compact placement design is described. It then describes the error in obtaining angular acceleration due to the phase and amplitude differences among the accelerometer outputs. It then introduces a “lead-lag compensation” method to eliminate the differences and hence eliminates the error. At the end of the chapter, the development and performance evaluation of the sensing sub-system based on the compact design is presented.

6.1. Introduction

In Chapter 5, propositions for the placement of accelerometers to obtain the highest possible angular sensing resolution for a given space constraint are described. As have been seen, sensing resolution of angular acceleration about the instrument axis is proportional to the width of the proximal-end sensor housing of the instrument. If the width of the proximal-end sensor housing of the instrument is small, the sensing resolution is low. Therefore, the width of the sensor housing at the proximal-end of the instrument cannot be made too slim for the sake of sensing resolution of angular motion about the instrument axis [Ang 03; Latt 09]. The size of the proximal-end sensor housing of the instrument, MicronII, in comparison with the hand holding the instrument can be seen in Fig. 6.1.

Fig. 6.1. A prototype of tremor compensation instrument, MicronII, with sensing sub-system following the proposed design II accelerometer placement.

It is discussed in [Patkin 77] that instruments that are heavy or stiff, and thick-handled would be clumsier and increase tremor. Although no discussion is made on the bulkiness of the instrument, it is apparent that bulky instruments will be more disruptive to surgeons/users than compact instruments due to the lack of mobility and potential blockage of user's view to the target by the instrument itself.

The greater the loading on the fingertip by weights, the more physiological tremor is increased [Patkin 77]. More weight at the end of the instrument will result in more loading on the fingertip because thumb cleft serves as a fulcrum point of the instrument. For that reason, possible adding of weight at the proximal end of the instrument should be avoided in designing instruments. On the other hand, as described previously, the width of the proximal-end sensing unit cannot be made too small and hence there is a compromise between the resolution and the ergonomics.

If angular acceleration about instrument axis α_z is not required in the calculation of tip acceleration vector, the proximal-end sensing unit can be made smaller and hence the problems associated with the bulkiness and weight of the instrument can be eliminated. In previous designs of sensing sub-system of tremor compensation instruments, which are described in Chapter 5, complete vector information on angular acceleration of the instrument is required to obtain complete vector information on the acceleration of the tip. However, if the placement of accelerometers is carefully considered, the information on angular acceleration about

Z_B axis, α_z , may not be required or no angular acceleration component may be required at all. For the reasons stated above, placement of accelerometers is analyzed and a placement design which complies with the propositions described in Chapter 5 and does not require information on angular acceleration about Z_B axis is proposed.

6.2. Compact Placement Design

Since there are three components in the acceleration vector, at least 3 accelerometers each sensing in different principal axis direction is required. Assuming accelerometers at location $\{1\}$, $\{2\}$, and $\{3\}$ sense accelerations in X , Y , and Z principal axis direction respectively, and using (4.8.a-4.8.c), acceleration at the instrument tip

$A_e = \begin{bmatrix} a_{e_x} & a_{e_y} & a_{e_z} \end{bmatrix}^T$ is found to be as follows.

$$a_{e_x} = a_{1_x} + p_{1e_z}\alpha_y - p_{1e_y}\alpha_z \quad (6.1.a)$$

$$a_{e_y} = a_{2_y} + p_{2e_x}\alpha_z - p_{2e_z}\alpha_x \quad (6.1.b)$$

$$a_{e_z} = a_{3_z} + p_{3e_y}\alpha_x - p_{3e_x}\alpha_y \quad (6.1.c)$$

where p_{ie_d} , $i=1,2,3$; $d=x,y,z$, is the displacement between location $\{i\}$ and the instrument tip along d principal axis. According to (6.1.a-6.1.c), apart from the three orthogonal acceleration readings from three accelerometers, three angular accelerations about three axes are required to get all the three acceleration components at the tip.

6.2.1. Sensing acceleration in the instrument axis direction, a_{e_z}

In this section, how the acceleration component in instrument axis direction, a_{e_z} , is obtained without requiring angular acceleration information is described. In (4.8.a-4.8.c),

$$\text{if } p_{ip_z} = p_{ip_y} = 0, a_{p_x} = a_{i_x}, \quad (6.2.a)$$

$$\text{if } p_{ip_x} = p_{ip_z} = 0, a_{p_y} = a_{i_y}, \quad (6.2.b)$$

$$\text{if } p_{ip_y} = p_{ip_x} = 0, a_{p_z} = a_{i_z} \quad (6.2.c)$$

The conditions for (6.2.a) and (6.2.c) are illustrated in Fig. 6.2. In the figure, acceleration component in Z_B axis direction at point R and that at point S are the same since $p_{RS_x} = p_{RS_y} = 0$. Similarly, acceleration component in X_B axis direction at point D is the same as that at point C since $p_{CD_z} = p_{CD_y} = 0$. Based on this concept, a sensing sub-system with lesser accelerometers is designed. In general, if accelerometers could be placed as shown in Fig. 6.3, only three accelerometers are all that is needed to obtain the complete vector information on acceleration of the tip.

Fig. 6.2. Illustration of the locations which have the same acceleration component in a particular axis direction (assuming angular velocity is small enough to be neglected).

In Fig. 6.3, accelerometers are placed on loci of Lines I, II, and III for sensing acceleration along X_B , Y_B , and Z_B axes respectively. For this configuration of accelerometer placement, no angular acceleration is required because $p_{1e_z} = p_{1e_y} = 0$ in (6.1.a) to calculate a_{e_x} , $p_{2e_x} = p_{2e_z} = 0$ in (6.1.b) to get a_{e_y} and $p_{3e_y} = p_{3e_x} = 0$ in (6.1.c) to get a_{e_z} .

Fig. 6.3. Theoretical placement of accelerometers to get a complete acceleration vector at the top without requiring angular motion information.

Although there is no problem in placing an accelerometer on the Line III, the instrument axis, to sense a_{e_z} , there are some implementation and ergonomic issues with placement of the other two accelerometers on Lines I and II and hence the two accelerometers are not placed on the Lines I and II while an accelerometer is placed on the instrument axis to sense a_{e_z} as shown in Fig. 6.4. From (6.1.c), since $p_{3e_y} = p_{3e_x} = 0$, $a_{e_z} = a_{3_z}$ meaning that the acceleration component at the tip in Z axis direction is just simply the accelerometer output.

Fig. 6.4. Placement of an accelerometer to sense acceleration component at the tip along Z_B axis direction.

6.2.2. Sensing acceleration components in the other two axes, a_{e_x} and a_{e_y}

Placing the other two accelerometers on the instrument axis for sensing acceleration in X_B axis, and Y_B axis directions as shown in Fig. 6.5 will eliminate the requirement of information on angular acceleration about Z_B axis since $p_{1e_y} = 0$ for the calculation of a_{e_x} in (6.1.a) and $p_{2e_x} = 0$ for a_{e_y} in (6.1.b).

Fig. 6.5. Placement of two accelerometers which require only two angular motion information to (α_y and α_x) obtain acceleration components in X_B axis and Y_B axis directions.

With the placement shown in Fig. 6.5, acceleration at the tip in Y_B axis, a_{e_y} can be calculated as

$$a_{e_y} = a_{2y} - \alpha_x p_{2e_z} \quad (6.3)$$

Acceleration at the tip in X_B axis direction can be calculated as

$$a_{e_x} = a_{1x} + \alpha_y p_{1e_z} \quad (6.4)$$

6.2.3. Requirement to obtain complete vector information on acceleration of the tip

Combining the placement of accelerometers in Fig. 6.4 and Fig. 6.5, complete vector information on acceleration of the tip is obtained. As can be seen from Fig. 6.4 and Fig. 6.5, obtaining complete vector information on the acceleration of the tip requires at least three accelerometers sensing along three principal axes, and angular

accelerations about X_B and Y_B axes (α_x and α_y). To get required angular accelerations, accelerometers are placed as shown in Fig. 6.6. The placement of accelerometers follows the propositions described in Chapter 5 to obtain highest possible angular sensing resolution.

Fig. 6.6. Placement of accelerometers to get a complete acceleration vector at the tip.

The calculations of α_x and α_y are simply as follows. Equation (4.4) is reproduced here. $A_{12} = A_2 - A_1 = \Omega \times (\Omega \times R_{12}) + \alpha \times R_{12}$. Again, in micromanipulation tasks, angular velocity is almost always very small. Hence, (4.4) becomes

$$A_{12} = A_2 - A_1 = \alpha \times R_{12}. \quad (6.5)$$

In components form, (6.5) becomes:

$$\begin{bmatrix} a_{12_x} \\ a_{12_y} \\ a_{12_z} \end{bmatrix} = \begin{bmatrix} \alpha_x \\ \alpha_y \\ \alpha_z \end{bmatrix} \times \begin{bmatrix} p_{12_x} \\ p_{12_y} \\ p_{12_z} \end{bmatrix} = \begin{bmatrix} \alpha_y p_{12_z} - \alpha_z p_{12_y} \\ -(\alpha_x p_{12_z} - \alpha_z p_{12_x}) \\ \alpha_x p_{12_y} - \alpha_y p_{12_x} \end{bmatrix}$$

Since, $p_{12_x} = p_{12_y} = 0$,

$$\begin{bmatrix} a_{12_x} \\ a_{12_y} \\ a_{12_z} \end{bmatrix} = \begin{bmatrix} \alpha_y p_{12_z} \\ -\alpha_x p_{12_z} \\ 0 \end{bmatrix} \quad (6.6)$$

$$\alpha_y = \frac{a_{12_x}}{p_{12_z}} = \frac{a_{2_x} - a_{1_x}}{p_{12_z}} \quad (6.7)$$

$$\alpha_x = -\frac{a_{12_y}}{p_{12_z}} = \frac{a_{1_y} - a_{2_y}}{p_{12_z}} \quad (6.8)$$

6.2.4. Obtaining complete vector information on acceleration of the tip

With the proposed placement of accelerometers as shown in Fig. 6.6, and using (4.8), tip acceleration is calculated in general as follows.

$$a_{e_x} = a_{i_x} + p_{ie_z} \alpha_y - p_{ie_y} \alpha_z, i = 1 \text{ or } 2. \quad (6.9)$$

$$a_{e_y} = a_{i_y} + p_{ie_x} \alpha_z - p_{ie_z} \alpha_x, i = 1 \text{ or } 2. \quad (6.10)$$

$$a_{e_z} = a_{3_z} + p_{3e_y} \alpha_x - p_{3e_x} \alpha_y \quad (6.11)$$

Since $p_{1e_x} = p_{1e_y} = p_{2e_x} = p_{2e_y} = p_{3e_x} = p_{3e_y} = 0$, equations (6.9-6.11) become

$$a_{e_x} = a_{i_x} + p_{ie_z} \alpha_y, i = 1 \text{ or } 2. \quad (6.12)$$

$$a_{e_y} = a_{i_y} - p_{ie_z} \alpha_x, i = 1 \text{ or } 2. \quad (6.13)$$

$$a_{e_z} = a_{3_z} \quad (6.14)$$

6.3. Elimination of Phase and Attenuation Differences among Accelerometer Outputs for Accurate Angular Sensing

Sensors or transducers convert one form of energy such as light, displacement, etc to another form such as voltage, current, etc that is suitable for a computer system to be acquired in. Electrical circuits are often employed in the conversion process. Depending on the type of circuits, some of them inevitably change the characteristics of the energy such as magnitude and phase.

For example, ADXL accelerometers, which are employed in the sensing sub-system, have output circuits which change the original acceleration signal's magnitude and phase. If output circuits of two accelerometers experiencing the same acceleration have same transfer function, the acceleration outputs of the circuits will have the same magnitude and phase characteristics. On the other hand, if the transfer functions of the circuits are different, one of the signals' phase will be leading the other and signal's

magnitudes will be different although the accelerometers are experiencing the same acceleration.

There is an internal resistor (R_{Filt}) at each output inside the integrated circuit (IC) chip as shown in Fig. 6.7.

Fig. 6.7. Functional block diagram of ADXL203 accelerometer [Analog 08].

A capacitor (C) has to be added at each output pin (X_{OUT} pin and Y_{OUT} pin in the figure) for the operation. The combination of an internal resistor and a capacitor forms a first order low-pass filter with a time-constant of $R_{Filt}C$. Transfer function of the filter can be expressed as:

$$H(s) = \frac{1}{1 + \tau s} \quad (6.15)$$

where τ is the time-constant of the filter.

6.3.1. The effect of the tolerance of internal resistor on phase and magnitude

Since resistance value of an internal resistor at each output circuit can vary typically as much as $\pm 25\%$ of its nominal value ($32 \text{ k}\Omega$) [Analog 08], time constant ($\tau = R_{Filt}C$) varies accordingly. Maximum value of R_{Filt} , $\max(R_{Filt})$ is $32\text{k} + 0.25(32\text{k}) = 40\text{k}$. Minimum value $\min(R_{Filt})$ is $32\text{k} - 0.25(32\text{k}) = 24\text{k}$. Using the capacitance value of $0.1 \text{ }\mu\text{F}$, the maximum time constant $\tau_{\max}$ is $\tau_{\max} = \max(R_{Filt})C = 4\text{ms}$, and the minimum time constant is

$\tau_{\min} = \min(R_{Filt})C = 2.4ms$. Therefore, the difference between the time constants of two accelerometer outputs can be as large as 1.6 ms.

The phase response of the RC low-pass filter can be written as:

$$\phi(j\omega) = -\tan^{-1}(\omega\tau) \quad (6.16)$$

The phase difference due to the difference in time-constants of 1.6 ms for a 10 Hz signal is about 5.5 deg as shown in the following equation and in Fig. 6.8.

$$\begin{aligned} \phi_{diff} &= -\tan^{-1}(2\pi \times 10 \times 4m) - (-\tan^{-1}(2\pi \times 10 \times 2.4m)) \\ &= -14.11 - (-8.5) = -5.53 \text{ deg} \end{aligned} \quad (6.17)$$

The magnitude response of the filter can be written as:

$$|G(j\omega)| = \frac{1}{\sqrt{1 + \omega^2 \tau^2}} \quad (6.18)$$

The difference in magnitude response due to the difference in the time constants $|G|_{diff}$ is $98.88 - 96.98 = 1.9 \%$ as shown below and in Fig. 6.8.

$$\begin{aligned} |G|_{diff} &= \frac{1}{\sqrt{1 + (2\pi \times 10)^2 (2.4 \times 10^{-3})^2}} - \frac{1}{\sqrt{1 + (2\pi \times 10)^2 (4 \times 10^{-3})^2}} \\ &= 0.9888 - 0.9698 = 0.019 \end{aligned} \quad (6.19)$$

Therefore, the total difference between two acceleration outputs due to the difference in phase and magnitude can be as large as about 10 % of the signal as can be seen in Fig. 6.9.

Fig. 6.8. Magnitude and phase responses of a first order low-pass filter at 10 Hz input frequency with different time constant values.

Fig. 6.9. The outputs of first order low-pass filters with time constants of 2.4 ms and 4 ms for 10 Hz signal input, and the difference between the two outputs.

6.3.2. *Effect of differences in phase-shift and magnitude on angular acceleration*

As can be seen from equations (7.7) and (7.8), two accelerometer outputs are required to calculate angular acceleration about a particular axis. Only when there is no difference in the amount of phase-shift and attenuation in the outputs, will calculation of the angular acceleration be accurate. Since the accelerometer outputs have different amounts of phase-shift and attenuation, the calculated angular acceleration will not be accurate without elimination of the differences in the amount of phase-shift and attenuation among accelerometer outputs.

Since the difference between two acceleration outputs can be as large as 10% of the outputs, angular acceleration error will consequently be 10% of the actual angular acceleration. It is described in section 5.6 that the maximum amplitude of angular acceleration of the instrument due to tremor about 4 rad/s^2 . Therefore, maximum angular acceleration error can be 0.4 rad/s^2 . Again, as discussed in section 5.7.2, multiplication of the displacement of 100 mm with angular acceleration is required to obtain the tip acceleration. The error in sensing of tip acceleration due to the angular acceleration error becomes 40 mm/s^2 which is equivalent to about $20 \text{ }\mu\text{m}$ p-p for a 10 Hz tremor signal (derivation to obtain $20 \text{ }\mu\text{m}$ can be referred back to section 5.7.2). Since the error is quite significant, it is of paramount importance that the source of the error, the differences in phase-shift and magnitude among accelerometer outputs, be eliminated.

6.3.3. *Lead-lag Compensation*

To compensate for the phase-shift and attenuation, a lead-lag compensator is proposed and employed for each output of the accelerometers. The transfer function of a first-order hardware low-pass filter formed by a resistor and a capacitor at the output of accelerometer $i, i = 1, 2, \dots, n$, is

$$H_i(s) = \frac{1}{1 + \tau_i s} \quad (6.20)$$

where $\tau_i = R_i C_i$ is time-constant of output circuit of accelerometer i .

The transfer function of a lead-lag compensator for the output of accelerometer i , $H_i^c(s)$ is then designed to be as follows:

$$H_i^c(s) = \frac{1 + \tau_i s}{1 + \tau s} \quad (6.21)$$

where τ is a time-constant for overall transfer function, $H_i^o(s)$ which then becomes

$$H_i^o(s) = H_i(s)H_i^c(s) = \frac{1}{1 + \tau_i s} \frac{1 + \tau_i s}{1 + \tau s} = \frac{1}{1 + \tau s} \quad (6.22)$$

Since $H_i^o(s)$ is independent of τ_i ,

$$H_1^o(s) = H_2^o(s) = \dots = H_n^o(s) = \frac{1}{1 + \tau s} \quad (6.23)$$

Block diagram of obtaining accurate angular accelerations using lead-lag compensators is shown in Fig. 6.10. The symbols a_i'' represent acceleration at the output of the compensator, while a_i are the actual acceleration at accelerometer i . Since all the a_i'' are the outputs of the same transfer function, they all have the same phase-shift and magnitude, and hence the error due to phase-shift and magnitude difference is eliminated.

Fig. 6.10. Block diagram of obtaining accurate angular accelerations using lead-lag compensators.

One requirement of the proposed method is the need to know the time-constant values of the output circuits of accelerometers. In the next section, how the time-constant values are obtained is described.

Measurement of time-constants of output circuits of accelerometers

Self-test feature (ST pin) of the accelerometer is used to find out exact time-constant. The original purpose of ST pin for the self-test feature is to easily see whether the accelerometer is functional. When this pin is set to V_s , the supply voltage, an electrostatic force is exerted on the beam of the accelerometer. The resulting movement of the beam allows the user to test if the accelerometer is functional. This pin can be used to find out the time-constant required for the compensator. The time-constant is known by measuring the time taken for the step output of the accelerometer to reach 63.2% of its final value.

6.3.4. Simulation, results, and discussion

In order to verify that the proposed method works, a simulation is performed. For simplicity, it is assumed that the instrument, *ITrem*, experiences only translational acceleration in each principal axis of the instrument body frame. Therefore, all the accelerometers having the same orientation experience the same acceleration and hence angular acceleration should be zero if there is no error. The profile of translational acceleration in each principal axis is simulated as a representative of tremor signal as follows:

$$a_{i_d} = 200 \sin(2\pi \times 10 \times kT) \text{ mm/s}^2 \quad (6.24)$$

To simulate a real-world situation, Gaussian white noise of 10 mm/s^2 RMS is superimposed onto the simulated signal. The noise superimposed acceleration signals, a_{1_x} and a_{2_x} , are shown in Fig. 6.11 (a). The signals a_{1_x} and a_{2_x} are then filtered using first order low-pass filter having time-constants of 2.4 ms and 4 ms respectively. The outputs of the filters, a'_{1_x} and a'_{2_x} , are shown in Fig. 6.11(b). The filter outputs are then fed into compensators. The outputs of the compensators, a''_{1_x} and a''_{2_x} , are shown in Fig. 6.11(c).

Fig. 6.11. (a) Actual acceleration superimposed by Gaussian white noise, (b) acceleration outputs of accelerometer output circuits, and (c) acceleration outputs of lead-lag compensators. Time-constant used for overall transfer function is 3 ms.

To see the effect of using compensators on angular acceleration error, angular acceleration formula from (6.7) is used. Angular acceleration errors with and without using compensators are shown in Fig. 6.12. As can be seen in Fig. 6.12, angular acceleration errors are reduced substantially. The error reduction by using compensators is about 60 %. The remaining errors in Fig. 6.12 (b) are due to the noise superimposed onto the signal. Although the proposed method is simple, its use is essential since the errors would be substantial without it. Therefore, in subsequent calculation of angular acceleration using (6.7-6.8) and tip acceleration using (6.12-6.14), acceleration outputs after the elimination of phase and attenuation differences, a''_{i_d} will be employed.

Fig. 6.12. Angular acceleration errors (a) without, and (b) with compensators.

6.4. Experiments and Results

A compact version of the tremor compensation instrument, “*ITrem*”, is developed with the sensing sub-system as shown in Fig. 6.13. The placement of accelerometers in the sensing sub-system is according to the compact proposed design shown in Fig. 6.6. The comparison of the size of *ITrem* and that of MicronII whose sensing sub-system consists of the accelerometer placement according to proposed design II is shown in Fig. 6.14.

Fig. 6.13. Photo of the accelerometers inside *ITrem* without the actuators and mechanism at the distal end.

To confirm that the angular acceleration formulae described in (6.7) and (6.8) are valid, an experiment is conducted. *ITrem* is placed on a one degree-of-freedom

rotating arm as shown in Fig. 6.15. The distal end of the sensing sub-system is attached with a non-reflective black color rod. The tip of the rod is attached with a reflective ball whose location is sensed using M^2S^2 [Latt 08b; Win 07] which is described in Chapter 3. The location of the ball with respect to the center of rotation of the rotating arm is known.

(a)

(b)

Fig. 6.14. Comparison of the sizes of the instruments.

(a)

(b)

Fig. 6.15. A photo of (a) $ITrem$ attached to the rotating arm, and (b) close-up view of the instrument tip placed near to the workspace of M^2S^2 .

Rotational motion of the instrument is produced by manually shaking the rotating arm horizontally back-and-forth. The rotational acceleration of the instrument is calculated from the M^2S^2 measured positions of the ball which is attached to the instrument. The measured position of the ball is then double-differentiated to obtain the ball acceleration which is further divided by the distance from the location of the ball to the center of rotation to obtain angular acceleration. Since the angular acceleration obtained contains noise due to the differentiation, offline zero-phase band-pass filtering is performed using a 4th order band-pass filter having a pass-band of 7-13Hz. The filtered angular acceleration is used as a ground truth angular acceleration.

The comparison of ground truth angular acceleration about Y_B axis and the calculated angular acceleration about Y_B axis obtained using the formula (6.15) and the

acceleration readings from the accelerometers in *ITrem* is shown in Fig. 6.16. The calculated angular acceleration from *ITrem* is also filtered using the same band-pass filter used in obtaining the ground-truth angular acceleration.

Fig. 6.16. Comparison of ground truth angular acceleration about Y_B axis and calculated angular acceleration about Y_B axis using acceleration outputs from accelerometers inside the instrument.

To prove the kinematic equations described in (6.12-6.14) are correct, an experiment is conducted. The positions of the ball attached to the tip of the instrument are measured while the instrument is being held as shown in Fig. 6.17. The instrument is held in a manner that orientation of the instrument with respect to the M^2S^2 is known. Four trials are conducted with each trial lasting for 16 seconds. Tip positions and accelerations due to tremor obtained from *ITrem* and M^2S^2 are used for evaluation of *ITrem* sensing.

Tip positions in X_B , Y_B , and Z_B axes due to tremor from *ITrem* are obtained by double-integrating the accelerations obtained from (6.12-6.14) and filtering the results using a 4th order band-pass filter having a pass band of 7-13 Hz. Tip positions from M^2S^2 are also filtered using the same band-pass filter to extract the positions due to tremor and remove the low frequency motion and the noise.

Tip accelerations due to tremor from *ITrem* are obtained by filtering the accelerations obtained from (6.12-6.14) using the band-pass filter while those from M^2S^2 are obtained by double-differentiating the ball positions and filtering the results.

Experiment results for sensing evaluation of *ITrem* are shown in Table 6.1. The values shown are average values of the four trials. RMS (μm) and RMS (mm/s^2) respectively are root mean squared values of positions and accelerations of the tip due to tremor obtained from M^2S^2 . RMSE (μm) and RMSE (mm/s^2) are obtained by respectively subtracting the ground-truth positions and accelerations from the calculated positions and accelerations of the tip due to tremor obtained from *ITrem*. Positions and accelerations of the tip due to tremor obtained from M^2S^2 are used as ground-truth positions and accelerations respectively.

Sensing accuracies of *ITrem* are obtained as follows:

$$\text{Position accuracy (\%)} = \frac{(\text{RMS}(\mu\text{m}) - \text{RMSE}(\mu\text{m}))}{\text{RMS}(\mu\text{m})} \times 100$$

$$\text{Acceleration accuracy (\%)} = \frac{(\text{RMS}(\text{mm/s}^2) - \text{RMSE}(\text{mm/s}^2))}{\text{RMS}(\text{mm/s}^2)} \times 100$$

Fig. 6.17. A photo taken while tremor motion of the ball is being measured using M^2S^2 as well as accelerometers inside *ITrem* to evaluate *ITrem* sensing.

Table 6.1 Experiment results for sensing evaluation of *ITrem*.

	X axis	Y axis	Z axis
RMS (μm)	17.37	17.84	9.87
RMS (mm/s^2)	70.74	72.92	36.15
RMSE (μm)	2.44	2.52	1.04
RMSE (mm/s^2)	13.09	10.95	4.56
Position accuracy (%)	86.59	84.97	89.79
Acceleration accuracy (%)	81.75	84.87	87.41

Plots of comparison of positions of the tip due to tremor calculated from *ITrem* and those obtained from M^2S^2 are shown in Fig. 6.18.

Fig. 6.18. Plots of comparison of position calculated from *ITrem* and that measured from M^2S^2 .

6.5. Discussion

The comparison in Fig. 6.16 shows that the formulae in (6.7) and (6.8) are valid although there are some errors. The errors are probably due to the accuracy of the M^2S^2 and the noise crept into the acceleration obtained from double-differentiation of the measured ball position by M^2S^2 .

Plots in Fig. 6.18 and accuracies shown in Table 6.1 show that the kinematic formulae (6.12-6.14) are correct although there are some errors. The errors might have resulted from the combined errors of the inaccurate positioning and orientation of the accelerometers, inaccurate measurement of the distance from the tip to an accelerometer, noises in the accelerometer outputs, accuracy and resolution of the measurement of the ball position by M^2S^2 . The positioning and orientation errors of the accelerometers can be known by using high precision rotation and translation stages and they can be corrected. However, since the stages were not available to the author when this research was carried out, these errors could not be identified and eliminated. That means the errors could have been reduced.

Since the sensing sub-system cannot provide angular acceleration about Z_B axis, acceleration component in Z_B axis at locations which are not on the instrument axis cannot be calculated. This might restrict the use of the proposed compact sensing sub-system with some unusual tools whose contact point with the target does not lie on the instrument axis.

Many applications do not require the instrument to be equipped with those unusual tools. For example, cells manipulation, injection of liquid medicine into a vessel, etc require the instrument to be attached with needles or needle-like parts at the center of the instrument's distal end. In these applications, one of the goals is to hit the target accurately with the tip of the needle or needle-like part. In these applications, α_z is not required and this proposed compact sensing design should be used for avoiding problems associated with bulkiness and heaviness of the instrument.

Except for redundant components, the fewer the components an instrument has, the smaller the probability of the instrument being failure and hence the more reliable it is. Comparing to that in the previous designs [Ang 03; Latt 08], fewer number of accelerometers is employed in the proposed compact placement design and hence the compact placement design can also be considered more reliable.

6.6. Conclusion

The contributions in the chapter include design, development, and evaluation of a compact sensing sub-system of the instrument, and elimination of angular-motion error by employing a proposed simple yet elegant “lead-lag compensation” method. The compact sensing sub-system is superior in terms of ergonomics, and reliability. By using the proposed compensation method, significant error reduction is found. Although the method is applied to the outputs of accelerometers of the hand-held instrument, is universal since it can be used to eliminate the phase and magnitude differences among outputs of multiple sensors or sensor modules as long as their time constants are known.

6.7. References

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Chapter 7 : Calibration of Accelerometer Orientation Errors

In this chapter, calibration of orientation error which exists in the placement of accelerometers is described. To perform calibration, orientation errors have to be identified first. Methods of identification of orientation error using gravity and the difficulties with the methods are described. A better method of identification using external orthogonal acceleration is then proposed, describing the advantages of the method. Online calibration or compensation for the errors is presented. The proposed method is evaluated with an experiment and simulation.

7.1. Introduction

Orientation errors exist in the placement of accelerometers. Orientation errors occur mainly at the system level, arising from machining errors of the mechanical structure or the body on which the sensors are mounted and/or assembly errors of the accelerometers to the mechanical structure (see Fig. 7.1 (a)). Orientation errors also result from packaging alignment error of accelerometer sensing axes in an integrated circuit (IC) package [Analog 08] (see Fig. 7.1 (b)).

Fig. 7.1. Illustration of accelerometer orientation errors resulting from (a) machining error and/or assembly error of the accelerometer to the mechanical structure, and (b) accelerometer IC packaging error.

Due to the accelerometer orientation error, acceleration reading from an accelerometer does not represent acceleration in the body frame or in a particular

direction of interest or ideal sensing direction which is shown in Fig. 7.2. Taking the acceleration reading as acceleration in the body frame or in the direction of interest affects accuracy of linear and angular acceleration estimation.

Orientation error is the misalignment between the accelerometer sensing axis (actual sensing direction) and the orthonormal principal body axes $X_B = [1 \ 0 \ 0]^T$, $Y_B = [0 \ 1 \ 0]^T$, and $Z_B = [0 \ 0 \ 1]^T$ (see Fig. 7.2). The directional cosines of actual sensing axes of the accelerometers with respect to the principal axes are defined as

$$\Theta_{ix} = [\cos \theta_{ix_x} \ \cos \theta_{ix_y} \ \cos \theta_{ix_z}]^T \quad (7.1)$$

$$\Theta_{iy} = [\cos \theta_{iy_x} \ \cos \theta_{iy_y} \ \cos \theta_{iy_z}]^T \quad (7.2)$$

$$\Theta_{iz} = [\cos \theta_{iz_x} \ \cos \theta_{iz_y} \ \cos \theta_{iz_z}]^T \quad (7.3)$$

where Θ_{ix} , Θ_{iy} , and Θ_{iz} , $i=1,2,\dots,n$ represent the directional cosines of actual sensing axes of the accelerometers at location $\{i\}$ with their ideal sensing directions being parallel to X_B , Y_B , and Z_B axis respectively. θ_{id_d} , $d=x,y,z$ is the angle between actual sensing axis and the orthonormal principal body axes where id represents the accelerometer at location $\{i\}$ with its ideal sensing direction being parallel to d principal body axis. The sub-subscript d represents one of the orthonormal principal body axes. The actual sensing axes of the accelerometers can be determined experimentally.

Fig. 7.2. Illustration of orientation error of an accelerometer. X_B , Y_B , and Z_B are orthonormal principal axes of the body frame attached to the body on which sensors are mounted.

A compensation method for acceleration estimation errors resulting from accelerometer orientation errors using angular velocity information is mentioned in [Tan 05]. Since angular velocity is obtained from angular acceleration by numerical integration, the error in angular velocity information grows linearly with time. Consequently, the error in misalignment correction grows linearly as time goes on. Therefore, the method has to rely on externally referenced sensing systems or sensors such as GPS to correct for the error. Since no externally referenced sensing systems or sensors are employed in the sensing sub-system of the instrument, the method is not applicable. Compensation for the acceleration estimation error due to the accelerometer orientation error is also shown in [Ang 04a] to obtain acceleration in the body frame or in ideal sensing direction without error. However, the method shown is only applicable to triaxial accelerometers. In some applications, single-axis or dual-axis accelerometers are preferred over the triaxial ones due to the high sensitivities. For example, dual-axis accelerometers are employed in the sensing sub-system of the instrument since they comparing to triaxial accelerometers have higher sensitivities which results in higher SNR. Therefore, a compensation method for the acceleration estimation error due to the accelerometer orientation error, which is applicable for any-axis accelerometer, is proposed.

7.2. Identification of Orientation Errors

This section describes methods of identification of accelerometer orientation errors using gravity and their drawbacks, and a better method of the identification is proposed.

7.2.1. Identification using gravity

A method of identifying orientation errors using gravity is described by Ang [Ang 04a]. In that method, the instrument has to be rotated until a particular accelerometer produces a maximum or minimum output.

To find Θ_{1z} , for example, the instrument is mounted with the Z_B axis of the instrument aligned with the gravity as shown in Fig. 7.3.

Fig. 7.3. Identification of accelerometer orientation error by finding maximum acceleration output from the accelerometer due to gravity, g .

The instrument is rotated about X_B and Y_B axes through the angles α° and β° (shown in Fig. 7.3) respectively until a maximum (or minimum if Z_B is aligned with $-g$) output voltage from a_{1z} is registered. The orientation of the real z-sensing axis a_{1z} with respect to the reference body frame is then found by

$$\Theta_{1z} = R_y(\beta_{1z})R_x(\alpha_{1z})Z_B = \begin{bmatrix} \cos \theta_{iz_x} & \cos \theta_{iz_y} & \cos \theta_{iz_z} \end{bmatrix}^T \quad (7.4)$$

where $R_x(\alpha_{1z})$ and $R_y(\beta_{1z})$ are 3×3 rotation matrices about X_B and Y_B with α_{1z}° and β_{1z}° amount of rotation respectively.

7.2.2. Sensitivity of detection of change in orientation

In this section, two main methods of identification of orientation errors in terms of sensitivity of detection of change in orientation of the accelerometer sensing axis are discussed.

The angle, γ , between actual sensing direction and ideal sensing direction due to misalignment is small and usually less than 3 deg. Packaging alignment error of ADXL-203 dual-axis accelerometers is stated as ± 1 deg [Analog 08]. Therefore, α and β values are also usually small.

Sensitivity when accelerometer ideal sensing axis is aligned with gravity

Supposing that the instrument is mounted with Z_B axis of the instrument aligned with the gravity as shown in Fig. 7.3, the acceleration reading from the accelerometer a_{1z} is as follows:

$$a_{1z} = g \cos \gamma \quad (7.5)$$

Let the value of γ be 1 deg, the value of a_{1z} becomes

$$a_{1z} = 0.9999g = 0.9999 \times 9810 = 9808.5 \text{ mm/s}^2$$

where g is gravitational acceleration.

When the actual sensing axis is perfectly aligned with the gravity, the value is the gravity value of 9810 mm/s^2 . The difference in acceleration reading for difference in angle of 1 deg is only 1.5 mm/s^2 .

Sensitivity when accelerometer ideal sensing is orthogonal to gravity

If the instrument is mounted with Z_B axis of the instrument orthogonal to the gravity, a_{1z} becomes

$$a_{1z} = g \cos(90 - \gamma) = g \sin \gamma = 0.0175g = 171 \text{ mm/s}^2$$

The acceleration reading is zero when the actual sensing axis is perfectly orthogonal to the gravity. The difference in acceleration reading for difference in angle of 1 deg is 171 mm/s^2 . Therefore, this method is 114 times ($171/1.5$) more sensitive than the method of finding the maximum or the minimum for 1 deg of orientation error.

Discussion on the above two methods

Noise always exists in sensor outputs. The presence of noise makes the method of finding the maximum or minimum acceleration values due to gravity difficult. The method of placing ideal sensing direction orthogonal to the gravity allows signal-to-noise ratio (SNR) to be higher (114 times higher for 1 deg of error), thereby obtaining more accurate misalignment error.

However, the later method requires the knowledge of bias voltage of accelerometer accurately. To know the bias voltage of an accelerometer, the maximum

and minimum acceleration values due to gravity have to be known, requiring the instrument to be rotated a number of times. To know a maximum voltage value of an accelerometer, a set of steps described in session 7.2.1 are required. For obtaining a minimum value, another set of the steps are needed. Therefore, for n number of accelerometers, $2n$ number of set of the steps is required, involving a number of rotations of the instrument. Moreover, the drift of the bias value with time [Ang 04b] makes identification of orientation error inaccurate.

7.2.3. Identification using external orthogonal acceleration

To identify orientation error of an accelerometer without the need to know the bias voltage, a method is proposed in which external translational accelerations acting on the instrument in the direction orthogonal to the accelerometer's ideal sensing direction are required. To find the orientation error of a_{iz} , for example, ${}^B a_{ix}$ and ${}^B a_{iy}$ are required as shown in Fig. 7.4.

Fig. 7.4. Identification of orientation error using external acceleration.

The external accelerations must be distinguishable from the bias. For example, it can be in sinusoidal profile with a known frequency. Since bias voltage is a DC offset voltage, it can be distinguished easily from the external acceleration using a linear filter or a Fourier Linear Combiner (FLC) [Vaz 94].

The external acceleration required can be created by moving the instrument in a known sinusoidal profile. Moving the instrument along each principal axis at a time, calculation of θ_{iz_d} , $d = x, y, z$, is obtained. Assuming external acceleration acting on

the location $\{i\}$ of the instrument in d axis of the instrument body frame be ${}^B a_{id}, d = x, y, z$, acceleration reading from an accelerometer at location $\{i\}$ with its ideal sensing direction being parallel to Z_B axis, a_{iz} is

$$a_{iz} = a_{iz}^x + a_{iz}^y + a_{iz}^z \quad (7.6)$$

where $a_{iz}^d, d = x, y, z$, are as follows:

$$a_{iz}^x = {}^B a_{ix} \cos \theta_{iz_x} \quad (7.7)$$

$$a_{iz}^y = {}^B a_{iy} \cos \theta_{iz_y} \quad (7.8)$$

$$a_{iz}^z = {}^B a_{iz} \cos \theta_{iz_z} \quad (7.9)$$

For example, moving the instrument along X_B axis, (7.7) is obtained. In (7.7), ${}^B a_{ix}$ is known input motion of the instrument, a_{iz}^x is acceleration reading from the accelerometer. Similarly, (7.8) and (7.9) are obtained. From (7.7) and (7.8), θ_{iz_x} and θ_{iz_y} are obtained as:

$$\theta_{iz_x} = \pm \arccos(a_{iz}^x / {}^B a_{ix}) \quad (7.10)$$

$$\theta_{iz_y} = \pm \arccos(a_{iz}^y / {}^B a_{iy}) \quad (7.11)$$

Equation (7.9) is not used to obtain θ_{iz_z} since it would give noisy value. The calculation of θ_{iz_z} is performed using results from (7.10) and (7.11) as follows:

$$\theta_{iz_z} = \pm \arccos(\cos(90 - \theta_{iz_x}) \cos(90 - \theta_{iz_y})) \text{ (or)} \quad (7.12-a)$$

$$\theta_{iz_z} = \pm [180 - \arccos(\cos(90 - \theta_{iz_x}) \cos(90 - \theta_{iz_y}))] \quad (7.12-b)$$

If the sign of a_{iz}^z is opposite to that of ${}^B a_{iz}$, the second solution (7.12-b) is chosen. The angles, θ_{ix_d} and $\theta_{iy_d}, d = x, y, z$, are found in a similar fashion.

The signs in (7.10), (7.11), (7.12-a), and (7.12-b) are chosen based on the prior-knowledge on the approximate orientation of the accelerometer sensing axis. For example, if the orientation of a particular accelerometer sensing axis is as shown in Fig.

7.4, then the approximate sensing direction is in the positive direction of the body frame and the positive sign will be taken.

7.3. Compensation for Acceleration Estimation Errors due to Accelerometer Orientation Errors

In this section, how acceleration in body frame or in ideal sensing direction is obtained from sensed acceleration from accelerometers is described.

7.3.1. Compensation for triaxial accelerometers

Equation (7.6) can be written using (7.7), (7.8), and (7.9) as:

$$a_{i_z} = \begin{bmatrix} \cos \theta_{iz_x} & \cos \theta_{iz_y} & \cos \theta_{iz_z} \end{bmatrix}^T A_i \text{ (or)}$$

$$a_{i_z} = \Theta_{iz}^T A_i \quad (7.13)$$

where $A_i = \begin{bmatrix} a_{ix} & a_{iy} & a_{iz} \end{bmatrix}^T$ which is to be determined from acceleration readings is acceleration in body frame at location $\{i\}$. It is also called body acceleration at $\{i\}$.

Similarly, a_{i_x} and a_{i_y} can be written as:

$$a_{i_x} = \Theta_{ix}^T A_i \quad (7.14)$$

$$a_{i_y} = \Theta_{iy}^T A_i \quad (7.15)$$

Combining (7.13-7.15), the following equation is obtained.

$$\begin{bmatrix} a_{i_x} & a_{i_y} & a_{i_z} \end{bmatrix}^T = \begin{bmatrix} \Theta_{ix} & \Theta_{iy} & \Theta_{iz} \end{bmatrix}^T A_i \quad (7.16)$$

$$A_i = M_i A_i \quad (7.17)$$

where $A_i = \begin{bmatrix} a_{i_x} & a_{i_y} & a_{i_z} \end{bmatrix}^T$, $M_i = \begin{bmatrix} \Theta_{ix} & \Theta_{iy} & \Theta_{iz} \end{bmatrix}^T$

$$A_i = M_i^{-1} A_i \quad (7.18)$$

Since matrix M_i is not orthonormal, $M_i^{-1} \neq M_i^T$. However, M_i^{-1} can be pre-computed and hence real time computation is not effected.

7.3.2. Compensation for non-triaxial accelerometers using triaxial accelerometer information

If accelerometers used include dual-axis accelerometers such as those in the sensing sub-system of the instrument or single-axis ones, one or more acceleration components are missing at each of the non-triaxial accelerometer location and (7.18) cannot be realized. However, a missing acceleration component whose ideal sensing direction being parallel to a particular body principal axis can be obtained from a body acceleration component in the same axis available at another triaxial accelerometer location by rigid body kinematics.

$$A_i = {}^B A_j + \alpha \times P_{ji} + \Omega \times \Omega \times P_{ji} \quad (7.19)$$

where A_i and ${}^B A_j$ are the vectors of acceleration at location i and body acceleration at location j respectively, $\alpha = [\alpha_x \ \alpha_y \ \alpha_z]^T$ is angular acceleration vector, Ω is angular velocity vector, and P_{ji} is displacement vector from location j to i .

As mentioned in an earlier chapter, in micromanipulation tasks such as microsurgical operations, angular velocity is very small and centripetal acceleration term $\Omega \times \Omega \times P_{ji}$ in (7.19) can be ignored.

$$A_i = {}^B A_j + \alpha \times P_{ji} \quad (7.20)$$

In component form, (7.20) can be written as

$$\begin{bmatrix} a_{i_x} \\ a_{i_y} \\ a_{i_z} \end{bmatrix} = \begin{bmatrix} {}^B a_{j_x} \\ {}^B a_{j_y} \\ {}^B a_{j_z} \end{bmatrix} + \begin{bmatrix} \alpha_y P_{ji_z} - \alpha_z P_{ji_y} \\ \alpha_z P_{ji_x} - \alpha_x P_{ji_z} \\ \alpha_x P_{ji_y} - \alpha_y P_{ji_x} \end{bmatrix} \quad (7.21)$$

For example if an acceleration component, a_{i_z} whose ideal sensing direction being parallel to Z_B axis is missing at accelerometer $\{i\}$, then (7.18) does not seem to be able to be solved since a_{i_z} in A_i and Θ_{i_z} in matrix M_i are not available. However, if there is a triaxial accelerometer at a location, at $\{j\}$ for example, as shown in Fig. 7.5, (7.18) can be solved since a_{i_z} can be known using (7.21) as follows:

$$a_{i_z} = {}^B a_{j_z} + \alpha_x P_{ji_y} - \alpha_y P_{ji_x} \quad (7.22)$$

where ${}^B a_{jz}$ is known using (7.18) and A_j . It is assumed here that α_x and α_y are known accurately.

Fig. 7.5. Illustration of missing acceleration component at a dual-axis accelerometer.

In M_i matrix in (7.18), Θ_{iz} is replaced with the unit vector $Z_B = [0 \ 0 \ 1]^T$. Then (7.18) becomes

$${}^B A_i = \left[\begin{bmatrix} \Theta_{ix} & \Theta_{iy} & Z_B \end{bmatrix}^T \right]^{-1} A_i \quad (7.23)$$

The reason for using Z_B as Θ_{iz} is that a_{iz} is obtained from ${}^B a_{jz}$ which is the acceleration in body frame obtained after compensation using acceleration readings at triaxial accelerometer. Therefore, ${}^B a_{jz}$ can be considered as acceleration reading from an accelerometer with no orientation error. Similarly, if y acceleration component is missing at location $\{i\}$, then Θ_{iy} in matrix M_i is replaced with the unit vector $Y_B = [0 \ 1 \ 0]^T$.

7.3.3. Iteration for better estimation

It is assumed that angular acceleration components required in the calculation of missing acceleration component using (7.21) are accurate. However, they are not accurate since they are calculated using acceleration readings at non-triaxial axis accelerometers because accelerations in body frame at these locations are not available at the time the missing acceleration components are calculated using (7.21). Consequently, the missing acceleration components and hence body accelerations at

non-triaxial accelerometer locations are not accurate. To improve the accuracy, iteration method is proposed.

At first, angular acceleration is calculated using acceleration readings. After the accelerations in body frame are obtained using the calculated angular acceleration for the first time, angular acceleration vector is calculated again using the accelerations in body frame. Then missing acceleration components are obtained using (7.21) and the most updated angular acceleration vector. The process is iterated until the differences in body acceleration between k^{th} iteration and $(k-1)^{\text{th}}$ iteration are below a predefined threshold.

$${}^B A_i(k) - {}^B A_i(k-1) < \sigma \quad (7.24)$$

where k represents number of iteration and σ is the threshold. The flow chart of the iteration is shown in Fig. 7.6. For the sake of clearer understanding, a schematic diagram of placement of accelerometers, which is shown in Fig. 7.7 is used to explain the iteration process.

Fig. 7.6. Flow chart of the iteration to obtain accelerations in body fame.

Before the iteration process, acceleration readings from non-triaxial accelerometers are taken as body accelerations.

$${}^B a_{id} = a_{id} \quad (7.25)$$

where a_{id} is the acceleration reading from a non-triaxial accelerometer at $\{i\}$ with its ideal sensing axis being parallel to d axis of the body frame. Then, accelerations in body frame at the triaxial accelerometer are computed using (7.18) as follows:

$${}^B A_i = \begin{bmatrix} \Theta_{1x} & \Theta_{1y} & \Theta_{1z} \end{bmatrix}^T A_i \quad (7.26)$$

Angular accelerations about X_B , and Y_B principal axes are estimated using acceleration readings from accelerometers in the formulae (7.27) and (7.28), respectively.

$$\alpha_x = \frac{{}^B a_{1y} - {}^B a_{2y}}{d_{12}} \quad (7.27)$$

$$\alpha_y = \frac{{}^B a_{2x} - {}^B a_{1x}}{d_{12}} \quad (7.28)$$

Fig. 7.7. Placement of accelerometers used to explain the iteration process.

By referring to (7.21), body acceleration at accelerometer $\{1\}$, ${}^B a_{1y}$ is used to obtain missing acceleration component a_{3y} at $\{3\}$ in Y_B axis as follows:

$$a_{3y} = {}^B a_{1y} + \alpha_z p_{13x} - \alpha_x p_{13z} \quad (7.29)$$

Since $p_{31x} = 0$, (7.29) becomes

$$a_{3y} = {}^B a_{1y} - \alpha_x p_{13z} \quad (7.30)$$

Body acceleration ${}^B a_{1z}$ is used to obtain missing acceleration component a_{2z} at $\{2\}$ as follows:

$$a_{2z} = {}^B a_{1z} + \alpha_x p_{12y} - \alpha_y p_{12x} \quad (7.31)$$

Since $p_{12x} = p_{12y} = 0$, (7.31) becomes

$$a_{2z} = {}^B a_{1z} \quad (7.32)$$

Once missing acceleration components have been calculated, body accelerations at non-triaxial accelerometers are calculated as follows:

By replacing Θ_{2z} in M_2 with $Z_B = [0 \ 0 \ 1]^T$, body acceleration at $\{2\}$, ${}^B A_2$ is obtained as:

$${}^B A_2 = \left[\begin{bmatrix} \Theta_{2x} & \Theta_{2y} & Z_B \end{bmatrix}^T \right]^{-1} A_2 \quad (7.33)$$

By replacing Θ_{3y} in M_3 with $Y_B = [0 \ 1 \ 0]^T$, body acceleration at $\{3\}$, ${}^B A_3$ is obtained as:

$${}^B A_3 = \left[\begin{bmatrix} \Theta_{3x} & Y_B & \Theta_{3z} \end{bmatrix}^T \right]^{-1} A_3 \quad (7.34)$$

The steps from (7.27) to (7.34) are repeated until desired level of acceleration estimation accuracy is achieved.

7.3.4. Compensation with placement of accelerometers in *ITrem*

As mentioned, all the accelerometers employed in *ITrem* are dual-axis accelerometers. Although there is no triaxial accelerometer in the instrument, the above method of compensation for the acceleration estimation errors due to orientation errors can be carried out due to the placement of accelerometers in *ITrem*. The explanation is described below.

It has been shown in section 6.2.1 that in the rigid-body motions involving very small angular velocities such as those in micromanipulation tasks, acceleration in a particular direction at a point on a rigid body is the same as that in the same direction at another point on the same rigid body as long as the two points lie on the line which is parallel to the acceleration direction. This is again shown in Fig. 7.8. For example, the acceleration at point C in X_B axis direction is the same as that at point D in the same direction. Likewise, in Fig. 7.9 which shows the placement of accelerometers in *ITrem*,

acceleration at location {1} in Z_B axis direction is the same as those at location {2} and {3} in the same direction.

Fig. 7.8. Illustration of the locations which have the same acceleration component in a particular axis direction (assuming angular velocity is small enough to be neglected).

Since acceleration in Z_B axis direction can be known from a_{3z} at accelerometer {3}, acceleration in Z_B axis direction at {1} and {2}, missing components a_{1z} and a_{2z} respectively, can be known. However, since there exists orientation error in the placement of accelerometer sensing axis a_{3z} , to minimize the error caused by angular motion-induced acceleration, the a_{3z} is used as a_{1z} in accelerometer at {1} to form an equivalent triaxial accelerometer at {1} rather than being used as a_{2z} in accelerometer {2} since the separation distance between {2} and {3} is larger than that between {1} and {3}. Once an equivalent triaxial accelerometer is formed, iteration process can be carried out.

Fig. 7.9. A schematic diagram of placement of accelerometers in *ITrem*.

7.4. Experiment

To show the effectiveness of the proposed identification method, a simulation as well as an experiment is conducted. As for the experiment, the tremor compensation instrument is placed on the nanopositioning system (P-561.3CD from Physik Instrumente) which is shown in Fig. 7.10. In this experiment, the amplitude and frequency of sinusoidal translational acceleration produced by the system for the identification of the orientation errors are 789.57 mm/s^2 (1579.14 mm/s^2 peak-to-peak) and 20 Hz respectively.

Fig. 7.10. The tremor compensation instrument placed on the nanopositioning system for identification of accelerometer orientation errors.

Fig. 7.11 shows the filtered acceleration reading (solid curve) from an accelerometer whose ideal sensing direction is parallel to the translational acceleration, and the filtered acceleration reading (dotted curve) from an accelerometer whose ideal sensing direction is orthogonal to the translational acceleration.

Fig. 7.11. The acceleration (solid curve) sensed by an accelerometer whose ideal sensing direction is parallel to the translational acceleration, and the acceleration sensed by an accelerometer (dotted curve) whose ideal sensing direction is orthogonal to the translational acceleration.

Acceleration readings from accelerometers are filtered using an FLC algorithm [Vaz 94] with a notch frequency of 20 Hz to extract accelerations due to the translational motion. After filtering with the FLC, peak-to-peak noise in the acceleration readings is reduced to 5 mm/s^2 . Therefore, the resolution of identification of the orientation error becomes

$$d\theta = \arcsin\left(\frac{5}{2 \times 789.57}\right) = 0.18 \text{ deg}$$

After the orientation errors have been identified, compensation for the acceleration estimation errors is performed using the iterative method and known translational motion of 400 mm/s^2 . The results of the iteration are shown in Fig. 7.12 and Fig. 7.13.

Fig. 7.12. Calculated angular acceleration.

Fig. 7.13. Calculated body acceleration at an accelerometer location.

7.5. Discussion

The magnitude of acceleration used is 6.2 times smaller than that of the gravity. In other words, SNR is 6.2 times smaller. However, since sensitivity of this method is more than a hundred times higher (114 times for 1 deg of error (see discussion on sensitivity in section II-B)), the net SNR of this method is about 18 times higher (114/6.2) than that of the method of finding the maximum or the minimum value.

Therefore, the proposed method has superior resolution of obtaining orientation error in the presence of noise.

If higher resolution of identification of orientation error is needed, the translation stage that can provide bigger amplitude of the translational acceleration should be used. The proposed method does not require any rotational movement of the instrument reducing a significant amount time and effort required to identify the error. By producing a translational acceleration having a certain known frequency, the need to know an accurate bias voltage is eliminated.

The compensation method described can also be used with single-axis accelerometers. As can be seen from Fig. 7.12 and Fig. 7.13, the values converge after about 3 iterations, suggesting that the method is suitable for real-time applications. If all the accelerometers employed are of triaxial type, the iteration steps are not necessary.

7.6. Conclusion

One of the contributions in this chapter comes from the superiority of the proposed orientation error identification method in terms of sensing resolution and ease of identification. Another contribution is the method of online compensation for the errors using the identified orientation errors and angular motion information. The compensation method is suitable for the real-time applications.

7.7. References

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Chapter 8 : Real-time Estimation and Prediction of Physiological Tremor

In this chapter, real-time estimation of physiological tremor (tremulous motion) from the sensed motion consisting of tremulous motion and non-tremulous motion is presented. At the beginning, it presents real-time filtering algorithms based on truncated Fourier-series namely Fourier Linear Combiner (FLC), Weighted-frequency Fourier Linear Combiner (WFLC), and Bandlimited Multiple Fourier Linear Combiner (BMFLC) to estimate periodic motion (tremulous motion) from the combination of the periodic and non-periodic motion (the tremulous and non-tremulous motion). It should be noted that in the followings, the phrases “periodic motion” and “tremulous motion” are used interchangeably since the algorithms are not limited to estimation of physiological to tremor only. Moreover, the word “signal” and “motion” are used interchangeably since they are not limited to motion only.

These algorithms have limitation in that they cannot recover the original tremulous motion from the sensed motion information which is delayed due to inherent hardware present in sensor modules such as a low-pass filter at each output of the accelerometers employed in the sensing sub-system. Moreover, the algorithms do not perform well if the sensed motion contains non-tremulous motion whose magnitude is too large comparing to that of the tremulous motion.

To recover the original (without delay) tremulous motion from the sensed information, modifications to the algorithms are proposed. The proposed modified-algorithms are tested with simulation as well as real data. Analytical integration required to obtain position information, which is needed by the manipulator sub-system for compensation, from acceleration without drift is described. Combination of analytical integration with the modified algorithms to obtain instantaneous position (without drift and phase-lag) of the tremulous motion is also described.

8.1. Introduction

Phase-shift is a major concern in real-time applications such as tremor compensation. Although linear filters are successful in extracting desired signals from a mixture of desired signals (tremulous motion in the case of real-time tremor

compensation) and undesired signals (non-tremulous motion in the case of real-time tremor compensation), the inherent phase-shift [Oppenheim 01] is a major drawback.

Owing to the rhythmic nature of the tremor, spectral analysis methods are quite suitable to study the tremor. Most of the spectral analysis methods are based on fast Fourier transform (FFT). The FFT is computationally inexpensive, allowing rapid data analysis and real-time implementation. The methods based on the FFT assume that the input signal is stationary. The input signal of finite-length is regarded as one cycle data of periodic sequence. However, since amplitude and frequency of the tremor are time-varying [Elble 90], the FFT-based methods are also not well-suited for extraction/estimation of tremor.

Time-frequency analysis methods such as short-time Fourier transform (STFT) are employed for many a biomedical applications. The STFT facilitates joint time-frequency analysis of the data. In STFT, the data is segmented into short period of sequences so that the data within the period of a segment can be assumed to be stationary. The shorter the period of sequences, the better the assumption of stationary, and the better the time resolution is. However, the price that is paid for the shorter period is that the frequency resolution is lower. Thus, there is a trade-off between time and frequency resolution in STFT.

To address the above issue, algorithms based on Fourier series such as Weighted-frequency Fourier Linear Combiner (WFLC) [Riviere 95a; Riviere 96; Riviere 98], and Bandlimited Multiple Fourier Linear Combiner (BMFLC) [Veluvolu 07a; Veluvolu 07b] are developed to estimate/extract desired periodic or quasi-periodic signals without causing changes in phase characteristics of the desired signals for real-time applications such as physiological and pathological tremor compensations [Riviere 95b; Riviere 03].

WFLC is an extension of the Fourier Linear Combiner (FLC) [Vaz 94] and can be considered as an adaptive zero-phase band-pass filter which can estimate a periodic or quasi-periodic signal consisting of a single dominant frequency and its harmonics. Proper performance of WFLC requires that the magnitude of undesired frequency components must not be large comparing to the desired frequency components. In the case of physiological and pathological tremor compensation, magnitudes of voluntary motion which is a component of non-tremulous motion are usually large with respect to the tremulous motion magnitude. One way to reduce the magnitude of the voluntary

motion which is a low-frequency motion is to employ a linear high-pass or band-pass filter. Although linear high-pass or band-pass prefiltering reduces the magnitude of undesired voluntary motion, it inherently introduces phase-shift. Since phase-shift is not desirable in real-time applications, a method is proposed in which WFLC-FLC combination [Riviere 95b] is employed. Ang *et al.* [Ang 06] introduces a method in which a combination of a low-pass filter and a high-pass filter by using frequency information provided by WFLC is employed.

One limitation of WFLC is its inability to extract a periodic or quasi-periodic signal containing more than one dominant frequency. To overcome the limitation, a new algorithm BMFLC [Veluvolu 07a; Veluvolu 07b] is proposed recently. Same as WFLC, BMFLC requires that the magnitude of undesired frequency components must not be large comparing to the desired frequency components. To reduce the magnitude of undesired frequency components without causing any change in phase characteristics of the original signal to be suitable for a BMFLC filter has been under research.

Although methods described in [Riviere 95b; Ang 06] do not introduce any phase-shift, they cannot compensate for a signal that has already had its phase and magnitude changed by inevitable filters such as a hardware filter which exists at the output of ADXL-203 accelerometers [Analog 08]. An innovative method of compensation for the phase-shift and attenuation incurred in the estimated periodic or quasi-periodic signals, which are caused by prefiltering prior to WFLC or BMFLC filters is presented in this chapter. The method is applicable to both WFLC and BMFLC filters and can even be used for prediction which is required in some applications.

The basic concept of the proposed method relies on the prior knowledge of the specifications of the linear filters employed in prefiltering and the frequency content of the desired periodic or quasi-periodic signal to be extracted. If the specifications of a filter and frequency of an input signal to the filter are known, the amount of attenuation and phase-shift of the signal at the output of the filter can be known. Using the knowledge on the amount of attenuation and phase-shift introduced by a particular filter for a particular frequency, compensation for the attenuation and phase-shift of each frequency component in the signal can be performed.

Since WFLC and BMFLC algorithms can provide information on the frequencies in the desired periodic or quasi-periodic signal together with their respective amplitudes, the algorithms are well suited for the proposed method.

8.2. Algorithms Based on Fourier Series

In this section, algorithms based on Fourier series such as FLC, WFLC, WFLC-FLC combination and BMFLC are described.

8.2.1. Fourier linear combiner (FLC)

A block diagram of FLC algorithm is shown in Fig. 8.1. Given a periodic signal with prior knowledge on its frequency, f_0 , the signal may be modeled in real-time by a Fourier Linear Combiner (FLC) [Vaz 94], which uses a least mean square (LMS) algorithm to adaptively adjust a series of Fourier components (w_{ik} , $i=1,2,\dots,2M$; in the figure).

Fig. 8.1. The FLC algorithm.

The reference input vector to FLC, $\vec{x}_k = [x_{1k} \dots x_{2Mk}]^T$ is:

$$x_{rk} = \begin{cases} \sin(r\omega_0 kT), r=1,2,\dots,M \\ \cos((r-M)\omega_0 kT), r=M+1,M+2,\dots,2M \end{cases} \quad (8.1)$$

where M is the number of harmonics used, $k=1, 2, \dots$ represents time-index, T is a sampling period. The LMS algorithm is used to update the weight vector as follows:

$$\varepsilon_k = s_k - \vec{w}_k^T \vec{x}_k \quad (8.2)$$

$$\vec{w}_{k+1} = \vec{w}_k + 2\mu \vec{x}_k \varepsilon_k \quad (8.3)$$

where $\bar{w}_k^T = [w_{1k} \dots w_{2M_k}]^T$ is the adaptive weight vector, s_k is the desired periodic or quasi-periodic signal which is to be modeled or estimated, μ is the adaptive gain parameter.

An estimation of the input signal can be calculated as:

$$\hat{s}_k = \bar{w}_k^T \cdot \bar{x}_k \quad (8.4)$$

8.2.2. Weighted-frequency Fourier linear combiner (WFLC)

If the frequency of the signal is unknown, WFLC [Riviere 95a; Riviere 96; Riviere 98], an extension to FLC, may be used. A block diagram of the WFLC algorithm is shown in Fig. 8.2.

The reference input vector to WFLC, $\bar{x}_k = [x_{1k} \dots x_{2M_k}]^T$ is:

$$x_{rk} = \begin{cases} \sin(rT \sum_{t=0}^k w_{ot}), r = 1, 2, \dots, M \\ \cos\left((r-M)T \sum_{t=0}^k w_{ot}\right), r = M+1, M+2, \dots, 2M \end{cases} \quad (8.5)$$

where M is the number of harmonics used, $k=1, 2, \dots$ represents time-index, T is a sampling period. As in FLC, weight vector is updated using LMS algorithm.

$$\varepsilon_k = s_k - \bar{w}_k^T \cdot \bar{x}_k \quad (8.6)$$

$$\bar{w}_{k+1} = \bar{w}_k + 2\mu \bar{x}_k \varepsilon_k \quad (8.7)$$

where $\bar{w}_k^T = [w_{1k} \dots w_{2M_k}]^T$ is the coefficient or weight vector of the reference input, s_k is the desired periodic or quasi-periodic signal which is to be modeled or estimated, and μ are the adaptive gain parameter.

The frequency, w_{0k} , which is required in the reference input vector, is estimated by modified LMS as follows:

$$w_{0_{k+1}} = w_{0k} + 2\mu_0 \varepsilon_k \sum_{i=1}^M i(w_{i_k} x_{M+i_k} - w_{M+i_k} x_{i_k}) \quad (8.8)$$

where μ_0 is adaptive gain parameter.

An estimation of the input signal can be calculated as:

$$\hat{s}_k = \vec{w}_k^T \cdot \vec{x}_k \quad (8.9)$$

Fig. 8.2. The WFLC algorithm.

8.2.3. WFLC-FLC combination

If the desired periodic or quasi-periodic signal is mixed with relatively large undesired frequency components such as noise, low-frequency components and DC offset, w_{0k} cannot adapt to the desired signal's frequency. To make the w_{0k} adapt well to the desired signal's frequency, WFLC-FLC combination [Riviere 98; Riviere 95a; Riviere 96; Riviere 95b] is proposed. The block diagram of the combination is shown in Fig. 8.3.

Fig. 8.3. WFLC-FLC combination.

The value of w_{0k} obtained from WFLC is used to compute the reference input $\bar{x}_k$. Using the computed reference input vector $\bar{x}_k$, the weights of the secondary FLC is updated via:

$$\hat{\varepsilon}_k = y_k - \bar{w}_k^T \bar{x}_k \quad (8.10)$$

$$\bar{w}_{k+1} = \bar{w}_k + 2\hat{\mu}\bar{x}_k \hat{\varepsilon}_k \quad (8.11)$$

where $\bar{w}_k^T$ is the coefficient or weight vector from the secondary FLC, $\hat{\mu}$ is the adaptive gain parameter for the FLC, y_k is the sensed signal. Thus, an estimation of the desired periodic or quasi-periodic signal can be calculated as:

$$\hat{s}_k = \bar{w}_k^T \bar{x}_k \quad (8.12)$$

8.2.4. Bandlimited multiple Fourier linear combiner (BMFLC)

One limitation of WFLC is its inability to extract a periodic or quasi-periodic signal containing more than one dominant frequency. To overcome that, an algorithm named BMFLC [Veluvolu 07b] is proposed. This approach relies on choosing a pre-determined band of frequencies based on the prior knowledge of the desired signal's frequency band. Spacing of frequencies is chosen according to user's requirement. The reference input to BMFLC is:

$$x_{r_k} = \begin{cases} \sin(2\pi f_r kT), r = 1, 2, \dots, N \\ \cos(2\pi f_{r-N} kT), r = N+1, N+2, \dots, 2N \end{cases} \quad (8.13)$$

where f_r are the frequencies within a given band of interest and N represents the number of frequencies used. The frequencies can be an integer as well as a rational number. The weights of BMFLC can be updated via:

$$\varepsilon_k = s_k - \vec{w}_k^T \cdot \vec{x}_k \quad (8.14)$$

$$\vec{w}_{k+1} = \vec{w}_k + 2\mu\vec{x}_k \varepsilon_k \quad (8.15)$$

An estimate of the desired signal can be given by:

$$\hat{s}_k = \vec{w}_k^T \cdot \vec{x}_k \quad (8.16)$$

Fig. 8.4. The BMFLC algorithm.

8.3. Compensation for Attenuation and Phase-shift

In this section, the proposed method of compensation for the attenuation and phase-shift incurred in the desired signal (periodic or quasi-periodic) to obtain its estimate accurately is described.

8.3.1. Compensation using WFLC

Let $m_{r_k}^l$ and $\phi_{r_k}^l$, $r=1,2,\dots,M$; $l=1,2,\dots,P$, respectively be the attenuation and phase-shift introduced by l^{th} filter for r^{th} harmonic frequency of the fundamental frequency w_{0_k} at k^{th} sample. Using WFLC, compensation for the attenuation and phase-shift is carried out by multiplying the coefficient vector with a modified reference input vector, $\bar{x}'_k = [x'_{1_k} \ x'_{2_k} \dots x'_{2M_k}]^T$ as follows:

$$x'_{r_k} = \begin{cases} \prod_{l=1}^P (1/m_{r_k}^l) \sin(rT \sum_{t=0}^k w_{0_t} - \sum_{l=1}^P \phi_{r_k}^l) \\ \text{for } r = 1, 2, \dots, M \\ \prod_{l=1}^P (1/m_{r-M_k}^l) \cos\left((r-M)T \sum_{t=0}^k w_{0_t} - \sum_{l=1}^P \phi_{r-M_k}^l\right) \\ \text{for } r = M+1, M+2, \dots, 2M \end{cases} \quad (8.17)$$

where P is the number of filters. A compensated estimate of the desired signal becomes

$$\hat{s}_k = \bar{w}_k^T \bar{x}'_k \quad (8.18)$$

The block diagram of the compensation using the modified reference input in WFLC is shown in Fig. 8.5.

8.3.2. Compensation using BMFLC

In the case of BMFLC, assuming m_r^l and ϕ_r^l , $r=1\dots N$ $l=1,\dots,F$ respectively are the attenuation and phase-shift introduced by l^{th} filter for f_r frequency, the modified input vector for compensation is as follows:

$$x'_{r_k} = \begin{cases} \left\{ \prod_{l=1}^F \frac{1}{m_r^l} \right\} \sin(2\pi f_r kT - \sum_{l=1}^F \phi_r^l), \\ r = 1, 2, \dots, N \\ \left\{ \prod_{l=1}^F \frac{1}{m_{r-N}^l} \right\} \cos(2\pi f_{r-N} kT - \sum_{l=1}^F \phi_{r-N}^l), \\ r = N+1, N+2, \dots, 2N \end{cases} \quad (8.19)$$

where F is the number of filters. A compensated estimate of the desired signal becomes

$$\hat{s}_k = \bar{w}_k^T \bar{x}'_k \quad (8.20)$$

The block diagram of the compensation using the modified reference input in BMFLC is shown in Fig. 8.6.

Fig. 8.5. Block diagram of the compensation using the modified reference input in WFLC.

Fig. 8.6. Block diagram showing how compensated estimate is obtained using BMFLC parameters.

A general block diagram of how the desired periodic or quasi-periodic signal is obtained from an attenuated and phase-shifted periodic or quasi-periodic signal is shown in Fig. 8.7.

Fig. 8.7. General block diagram showing how the desired signal is recovered from an attenuated and phase-shifted periodic or quasi-periodic signal.

The input signal to WFLC or BMFLC, s'_k is an attenuated and phase-shifted version of the desired signal s_k (periodic or quasi-periodic) due to prefiltering by one or more filters. The WFLC or BMFLC operates on the s'_k using a reference input vector (from (8.1) for WFLC and from (8.9) for BMFLC) and produces coefficient or weight vector $\bar{w}_k^T$ and estimate of s'_k , $\hat{s}'_k$. To obtain the estimate of s_k , the weight vector is used to perform an inner product operation with a modified reference input vector obtained from (8.13) for WFLC and (8.15) for BMFLC.

8.4. Simulation and Results

In order to show that the method works, a simulation is performed. Filters employed in the simulation are a first order Butterworth low-pass filter (LPF) and a

second order Butterworth high-pass filter (HPF) with cutoff frequencies of 15 Hz and 1 Hz respectively. The block diagram of the steps involved in the simulation with BMFLC is shown in Fig. 8.8.

For simulation with BMFLC, the simulated sensed signal comprising of sinusoidal components having amplitudes of 200 mm/s² and frequencies of 10 Hz, 11 Hz, and 11.5 Hz is used as a desired periodic signal to be estimated. The desired signal is superimposed with Gaussian white noise having standard deviation of 10 mm/s² and a DC offset value of 5000 mm/s². The DC offset is to simulate a sensor bias voltage while noise represents sensor noise.

Frequency responses of the filters are pre-calculated and those in the range of 8 to 12 Hz are shown in Fig. 8.9 and Fig. 8.10. The following parameters for BMFLC are used: $N = 5$, $\mu = 0.03$ and pass-band range for the periodic signal is set from 10Hz to 12Hz. The modified reference input for the BMFLC is then:

$$x'_{rk} = \begin{cases} \left\{ \prod_{l=1}^2 \frac{1}{m_r^l} \right\} \sin(2\pi f_r kT - \sum_{l=1}^2 \phi_r^l), & r = 1 \dots 5; \\ \left\{ \prod_{l=1}^2 \frac{1}{m_{r-5}^l} \right\} \cos(2\pi f_{r-5} kT - \sum_{l=1}^2 \phi_{r-5}^l), & r = 6 \dots 10; \end{cases} \quad (8.21)$$

with $f_1 = 10, f_2 = 10.5, f_3 = 11, f_4 = 11.5, f_5 = 12$

The values of m_r^l and ϕ_r^l which are known from the frequency response are put in the above modified reference input. For example, modified reference input components for 10 Hz frequency x'_{1k} and x'_{6k} become:

$$x'_{1k} = \frac{1}{m_1^1} \times \frac{1}{m_1^2} \sin(2\pi(10)kT - (\phi_1^1 + \phi_1^2)) \quad (8.22)$$

$$x'_{6k} = \frac{1}{m_1^1} \times \frac{1}{m_1^2} \cos(2\pi(10)kT - (\phi_1^1 + \phi_1^2)) \quad (8.23)$$

where $m_1^1 = 0.8322, m_1^2 = 1, \phi_1^1 = -33.68^\circ$, and $\phi_1^2 = 8.127^\circ$ (see Fig. 8.9 and Fig. 8.10). Similarly, other components of the modified reference input for other frequencies are found.

Fig. 8.8. Block diagram of the steps involved in the simulation.

Fig. 8.9. Frequency response of the first order low-pass Butterworth filter with a cutoff frequency of 15 Hz.

Fig. 8.10. Frequency response of the second order high-pass Butterworth filter with a cutoff frequency of 1 Hz.

Fig. 8.11 (a) shows the signal consisting of a modulated signal which is the desired signal to be estimated, DC offset, and Gaussian white noise. Fig. 8.11 (b) shows the HPF filter output or BMFLC input (dashed line) and the modulated signal (solid

line). As can be seen from the figure, the output at the HPF filter, s'_k is an attenuated and phase-shifted version of the desired signal, s_k . Fig. 8.12 shows the comparison of uncompensated estimate ($\hat{s}'_k$) and compensated estimate ($\hat{s}_k$) of the modulated signal. Estimation error with and without prefiltering is also shown in the figure.

For comparison of the proposed method with WFLC-FLC combination, a sinusoidal signal having amplitude of 200 mm/s² and frequency of 10 Hz is used as a desired periodic signal. The signal is superimposed with Gaussian white noise having standard deviation of 10 mm/s² and a low-frequency sinusoidal signal with amplitude of 2000 mm/s² and frequency of 1Hz.

Fig. 8.11. (a) simulated sensed signal consisting of the modulated (desired) signal, noise, and DC offset, and (b) modulated (desired) signal consisting of 10 Hz, 11 Hz, and 11.5 Hz (solid line), and the HPF filter output or BMFLC input (dashed line).

Fig. 8.12. Uncompensated estimate and compensated estimate of the modulated signal provided by BMFLC.

The low-frequency signal is to simulate an undesired component in the sensed signal. The following parameters for WFLC-FLC combination are used: $M = 1$, $\mu = 0.03$, $\mu_0 = 5e-6$, $\hat{\mu} = 0.03$.

Filters employed in the simulation are a 6th order Butterworth low-pass filter (LPF) and a 6th order Butterworth high-pass filter with cutoff frequencies of 15 Hz and 6 Hz respectively. To make the comparison fair, the same filters are used for both prefiltering in the proposed method using WFLC and in the WFLC-FLC combination. Comparison of estimation error of the proposed method and that of the WFLC-FLC combination is shown in Fig. 8.13 and Table 8.1. The errors of the estimation in Table 8.1 are obtained by subtracting the estimates from the periodic signal.

Fig. 8.13. Comparison of estimation error of the proposed method and that of the WFLC-FLC combination.

Table 8.1 Root mean square errors (*RMSE*) of estimation of WFLC-FLC combination and the proposed method

	WFLC-FLC	Proposed method	Error reduction (%)
RMSE (mm/s ²)	14.55	1.14	92.18

8.5. Performance with Physiological Tremor

In the previous section, the effectiveness of the proposed compensation method is shown using pure periodic signals. In order to show that the method works well with the quasi-periodic signals such as physiological tremor signals since tremor is approximately rhythmic (quasi-periodic) signal [Elble 90], and can be employed in physiological tremor compensation/cancellation, the method is tested with real physiological tremor data. The tremor data is obtained from surgical instrument tip

motion data which is collected [Su 09] during micromanipulation tasks performed by subjects using a micro motion sensing system (M^2S^2) [Win 07]. The instrument tip motion data is filtered off-line using an off-line zero-phase band-pass filter having a pass-band of 5-15 Hz to obtain physiological tremor and remove non-tremulous components such as low-frequency intended motion, and sensor and measurement noise.

To simulate a hardware filter present at the output of ADXL-203 accelerometers employed in the tremor compensation instrument [Ang 03], the tremor data is filtered using a first-order software low-pass filter having the time-constant of 3 ms which is a typical time-constant value of the hardware filter at the output of ADXL-203 accelerometers. The filtered tremor data is passed through a BMFLC filter to estimate it. The estimated output of the BMFLC filter has a phase-lag without using the compensation method.

Fig. 8.14 shows the effectiveness of the compensation method with physiological tremor. A phase-lag can be seen in uncompensated estimate. The phase-lag is significantly reduced in the compensated estimate. Fig. 8.15 shows physiological tremor estimation errors with and without compensation. Table 8.2 and Table 8.3 respectively show statistics of root-mean squared errors (RMSE) and peak-peak errors (P2PE) of ten trials of estimation of different physiological tremor signals with and without compensation, and the percentage of error reduction which is obtained by comparing the estimation error using the proposed compensation method and that without using the method. RMSE reduction (%) is obtained as follows:

$$\text{RMSE reduction (\%)} = \frac{\text{RMSE}_{wo} - \text{RMSE}_w}{\text{RMSE}_{wo}} \times 100 \quad (8.24)$$

Peak-peak error reduction (%) is obtained as follows:

$$\text{P2PE reduction (\%)} = \frac{\text{P2PE}_{wo} - \text{P2PE}_w}{\text{P2PE}_{wo}} \times 100 \quad (8.25)$$

Fig. 8.14. A plot showing the physiological hand tremor (ground-truth) of a subject, uncompensated estimate, and compensated estimate of the tremor.

Fig. 8.15. Physiological tremor estimation errors with and without compensation.

Table 8.2 Mean and standard deviation of RMSE of ten trials of estimation of different physiological tremor signals with and without proposed compensation method

	RMSE without compensation RMSE _{wo} (μm)	RMSE with compensation RMSE _w (μm)	RMSE reduction (%)
Mean	3.46	2.31	30.17
Standard deviation	1.42	0.73	11.46

Table 8.3 Mean and standard deviation of peak-peak errors of ten trials of estimation of different physiological tremor signals with and without proposed compensation method

	Peak-peak error without compensation P2PE _{wo} (μm)	Peak-peak error with compensation P2PE _w (μm)	Peak-peak error reduction (%)
Mean	29.07	22.72	20.83
Standard deviation	7.92	7.36	17.55

8.6. Prediction

Prediction of the signal can also be performed by adding one more phase term in the sinusoidal function of the modified reference input. In BMFLC, the modified reference input for prediction then becomes:

$$x'_{r_k} = \begin{cases} \left\{ \prod_{l=1}^F \frac{1}{m_r^l} \right\} \sin(2\pi f_r kT - \sum_{l=1}^F \phi_r^l + 2\pi f_r T_p), & r = 1 \dots N \\ \left\{ \prod_{l=1}^F \frac{1}{m_{r-N}^l} \right\} \cos(2\pi f_{r-N} kT - \sum_{l=1}^F \phi_{r-N}^l + 2\pi f_{r-N} T_p), & r = N + 1 \dots 2N \end{cases} \tag{8.26}$$

where T_p is the time in future in second at which the state of the periodic or quasi-periodic signal is to be estimated.

In WFLC, the modified reference input for prediction becomes:

$$x'_{r_k} = \begin{cases} \prod_{l=1}^P (1/m_{k_r}^l) \sin(rT \sum_{t=0}^k w_{o_t} - \sum_{l=1}^P \phi_{k_r}^l + r w_{o_k} T_p) \\ \text{for } r = 1 \dots M \\ \prod_{l=1}^P (1/m_{k_{r-M}}^l) \times \\ \cos\left((r-M)T \sum_{t=0}^k w_{o_t} - \sum_{l=1}^P \phi_{k_{r-M}}^l + (r-M)w_{o_k} T_p\right) \\ \text{for } r = M+1 \dots 2M \end{cases} \quad (8.27)$$

8.7. Analytical Integration

It is a well-known fact that numerical integration of acceleration information from accelerometers to obtain position information inherently causes position errors to grow quadratically with time, which is commonly known as “position drift”. To eliminate the position drift in the estimation of periodic motion using accelerometers, analytical integration method is proposed using WFLC and BMFLC.

Using acceleration as an input, WFLC or BMFLC gives an acceleration weight vector $\vec{w}_k^T$ which can be converted to position weight vector $\vec{w}_{p_k}^T$ to obtain the position estimate.

In the case of WFLC, $\vec{w}_{p_k}^T = [w_{p_{1k}} \dots w_{p_{2Mk}}]^T$ can be obtained as follows:

$$w_{p_{r_k}} = \begin{cases} -w_{r_k} / \left(\frac{r w_{0_k}}{T}\right)^2, r = 1 \dots M \\ -w_{r_k} / \left(\frac{(r-M)w_{0_k}}{T}\right)^2, r = M+1 \dots 2M \end{cases} \quad (8.28)$$

In the case of BMFLC, $\vec{w}_{p_k}^T = [w_{p_{1k}} \dots w_{p_{2Nk}}]^T$ can be obtained as follows:

$$w_{p_{r_k}} = \begin{cases} -w_{r_k} / (2\pi f_r)^2, r = 1 \dots N \\ -w_{r_k} / (2\pi f_{r-N})^2, r = N+1 \dots 2N \end{cases} \quad (8.29)$$

The estimate of the position is then obtained as:

$$\hat{p}_k = \vec{w}_{p_k}^T \cdot \vec{x}_k \quad (8.30)$$

8.8. Analytical Integration with Compensation

In this section, the combination of analytical integration with the proposed compensation method to recover from acceleration information from accelerometers the non-attenuated and non-phase-shifted position which has no position drift is presented. Fig. 8.16 shows a block diagram of how to obtain estimate of the position due to periodic motion which is sensed by an accelerometer.

The input to WFLC or BMFLC is attenuated and phase-lagged acceleration. The acceleration weight vector $\tilde{w}_k^T$ provided by WFLC or BMFLC is converted to position weight vector $\tilde{w}_{p_k}^T$ which in turn is used in the dot-product operation with the modified reference input $\tilde{x}'_k$ to obtain estimate of the position as:

$$\hat{p}_k = \tilde{w}_{p_k}^T \cdot \tilde{x}'_k \quad (8.31)$$

Fig. 8.16. Analytical integration and compensation for attenuation and phase-shift.

Experiment results of sensing and filtering using the proposed analytical integration with compensation method is presented in section 9.5.

8.9. Discussion

Compensated estimate shown in Fig. 8.12 suggests that the proposed method can provide accurate estimate of the periodic signal. Estimation errors produced by the BMFLC with and without prefiltering shown in the same figure suggests that prefiltering should be performed to obtain accurate estimates of periodic signals even if the sensors or sensor modules do not introduce any attenuation or phase-shift to the original signal.

Comparison of estimation errors of the proposed method and that of the WFLC-FLC combination in Table 8.1 indicates that the proposed method yields much better accuracy of estimation. Results of testing with real physiological tremor data shows that the proposed method works well with quasi-periodic signals and can be employed in tremor compensation applications.

Prediction of the desired signals is necessary for the better performance in some applications such as tremor compensation/cancellation. For example, in hand-held physiological tremor compensation instruments, the tremulous motion of the instrument tip is compensated using a micromanipulator. It is important to know the exact current position of the tip. Once the instantaneous tip position due to tremor has been estimated, the tip position is no longer at the estimated position since the tip is moving continuously and sensors detecting the motion that has already occurred. This is the principal of causality. Therefore, prediction is necessary to obtain the exact current position of the tip. Moreover, the micromanipulator takes some amount of time to move the tool tip to the desired position. This delay causes tremor compensation inefficient without prediction.

With the proposed prediction method, the inefficiency due to manipulator delay time and the effect of causality can be reduced. Overall prediction time, T_p in (8.26) and (8.27), (the amount of time in future at which the position of the tip is to be estimated) can be decided trial-and-error.

8.10. Conclusion

Contributions in this chapter are as follows. Modifications to the filtering algorithms (WFLC, and BMFLC) are proposed so that original (without phase-lag)

tremulous motion can be inferred from the motion information that is obtained later due to the presence of inherent hardware filters. Obtaining the original motion is important for more effective tremor compensation. Effectiveness of the proposed modification has been shown with simulated as well as real physiological motion data. A way to perform prediction of the signals using the WFLC and the BMFLC filters has also been shown. Analytical integration using the described filtering algorithms is introduced so that notorious position drift is eliminated. Combination of analytical integration and the proposed modification of the algorithms to obtain instantaneous position (without drift and phase-lag) of the tremulous motion is presented.

As a general suggestion, if the periodic (tremulous) motion consists of only one dominant frequency, WFLC should be employed. If it consists of more than one dominant frequency, BMFLC should be employed.

8.11. References

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Chapter 9 : The Complete Instrument (*ITrem*)

In this chapter, integration of sensing sub-system with filtering sub-system to obtain tremulous position of the tool tip is described. Then, manipulation sub-system is briefly presented which is followed by the section on hardware integration of the sensing sub-system with the manipulation sub-system to form a compact physical instrument, *ITrem*. Finally, preliminary evaluations and results of the tremor compensation performance of the instrument are presented and discussed.

9.1. Integration of Sensing and Filtering Sub-systems

Overall block diagram of sensing and filtering sub-systems is shown in Fig. 9.1. Instrument motion is sensed using accelerometers inside the instrument. To equalize attenuation and phase differences among the outputs of the accelerometers, lead-lag compensators are employed as described in section 6.3. The equalized acceleration outputs are used to calculate accelerations in body frame using the prior-knowledge on the accelerometer orientation errors. Angular acceleration components are also calculated using the accelerations in body frame. To remove gravity, and low-frequency components of acceleration, accelerations in body-frame are filtered. Then kinematics is used to compute acceleration of the tool tip.

The tremulous acceleration of the tip is estimated using BMFLC filter. As it is mentioned in Chapter 8, if more than one dominant frequency is present in the tremor, WFLC should be used. In that case, the block in yellow color will be replaced as shown in Fig. 9.2. Estimate of the tremulous position of the tool tip is then obtained by analytically double-integrating the estimate of the tremulous acceleration.

*i – More information about this block can be seen in chapter i.

Fig. 9.1. Overall block diagram of sensing sub-system and filtering sub-system with BMFLC filter.

Fig. 9.2. Block diagram of filtering sub-system with WFLC filter.

9.2. Manipulation Sub-system

This manipulation sub-system is designed by the author's fellow student, Tan U-Xuan. To compensate for the tremulous motion, a piezoelectric driven manipulator is designed and built. Piezoelectric actuators are employed because of its ability to output a large force, compactness in size, large operating bandwidth and rapid response. In order to perform tracking control of the actuator, Prandtl-Ishlinskii (PI) hysteresis model is used. Details of the controller design can be found in [Ang 07; Tan 08]. Instead of going for the usual serial mechanism, a parallel mechanism design is the preferred choice because of its rigidity, compactness and simplicity in design. To avoid backlash problem and for better resolution, flexure joints are used. To save cost, Objet, a rapid-prototyping machine, is proposed to be used. The design of the mechanism of the manipulator is shown in Fig. 9.3. Manipulation sub-system is also evaluated using M^2S^2 as shown in Fig. 9.4. Desired motions of the instrument tip are generated by the manipulator and the actual motions are recorded with M^2S^2 . A 3-D plot of paths of the end effector (the ball) when it is driven to move along a circular path is shown in Fig. 9.5. A 2-D plot of paths of the end effector (the ball) when driven to move along x-axis is shown in Fig. 9.6.

Fig. 9.3. Design of the mechanism of the manipulation sub-system [Tan 09].

Fig. 9.4. Experiment setup for the evaluation of manipulation sub-system performance.

It has been reported in [Tan 09] that RMSE of the manipulation sub-system is about 7 μm for the sinusoidal motion of ball at 85 μm amplitude and 10 Hz. This can be translated as 11% of the manipulation sub-system error. The range of the manipulator in x-axis and y-axis is about 170 μm .

Fig. 9.5. A 3-D plot of the end effector when driven to move along a circular path [Tan 09].

Fig. 9.6. A 2-D plot of the end effector when driven to move along x-axis [Tan 09].

9.3. Hardware Integration of the Sensing Sub-system with the Manipulation Sub-system

It is of importance that the instrument size is not big and its length including the surgical tool is less than 200 mm since the maximum height of the workspace under a surgical microscope is 200 mm. The reduction in the width of the proximal-end of the instrument is achieved using the placement design described in Fig. 6.6 on page 120. The reduction in length of the instrument results from the smaller design of the in-house accelerometer module which is shown in Fig. 9.7, and the integration design of the sensing sub-system and the manipulation sub-system as shown in Fig. 9.9.

Fig. 9.7. Comparison of sizes of an in-house accelerometer module and an off-the-shelf accelerometer module from Spark Fun Electronics.

Fig. 9.8. A mechanical drawing of integration of sensing and manipulation sub-systems.

Fig. 9.9. Integration of sensing and manipulation sub-systems: (a) distal-end accelerometer fixed to the front part of the actuator holder parts before the two parts are assembled together, (b) after the parts are assembled together, (c) and (d) the actuator holder with the mechanism attached.

The distal-end accelerometer is placed among piezoelectric actuators and near to the mechanism so that the length of the instrument can be shortened while the separation distance between the distal-end and the proximal-end accelerometers is kept the same as that in Micron. Instead of making one whole piece for holding actuators, two pieces are made so that the distal-end accelerometer board can be fixed among actuators as shown in Fig. 9.9 (a) and (b).

9.4. Overall System

Overview of the hand-held instrument, *ITrem* with active tremor compensation is shown in Fig. 9.10. The overall block diagram of *ITrem* is shown in Fig. 9.11.

Acceleration output from each accelerometer is acquired at 25 kSamples/s with a 16-Bit, 250 kSamples/s multifunction data acquisition card (NI PXI-6232, National Instruments) placed inside a PXI computer (NI PXI-8106, National Instruments). LabVIEW RT is used as a real-time operating system. The control loop time is set at 2 ms. Number of samples acquired from each accelerometer output in one control loop is therefore $25 \text{ kSamples} \times 2 \text{ ms} = 50 \text{ Samples}$. To reduce noise, the samples are then averaged and the averaged value is used in the control loop.

The sensing sub-system and the filtering sub-system provide information on the tool tip position due to physiological tremor. The information is used by the manipulation sub-system to calculate required voltages for the actuators. The scaled-down voltages are outputted through an analog output card (NI PXI-6733, National Instruments). The scaled-down voltages are then amplified using a power amplifier to obtain the required voltage levels for the actuators to actuate the mechanism which in turn manipulates the tool.

A photo of *ITrem* in comparison with a ruler is shown in Fig. 9.12. The instrument measures 189 mm in length (inclusive of the 55 mm long intraocular surgical needle), $\varnothing 22 \text{ mm}$ in the thickest portion and weighs 75g.

Fig. 9.10. Overview of the handheld instrument, *ITrem*, with active tremor compensation.

Fig. 9.11. Overall block diagram of *ITrem*.

Fig. 9.12. A photo of *ITrem* in comparison with a ruler.

9.5. Experiments and results

To evaluate the performance of the integration of sensing and filtering sub-systems as well as the tremor compensation performance of the instrument, bench tests and hand-held tests are conducted. In the bench test, the instrument is mounted on a nanopositioning system (P-561.3CD, Physik Instrumente, Germany) as shown in Fig. 9.13. The system is driven in X and Y axes to replicate real hand tremor movement of a surgeon recorded in one of the earlier experiments [Su 09] using M^2S^2 . The positioning error of the system is less than 100 nm.

Fig. 9.13. Setup for the bench test.

9.5.1. The performance of integration of sensing and filtering

Plots of ground-truth tremulous motion and the estimated tremulous motion of the tool tip are shown in Fig. 9.14. The motion of the nanopositioning system is used as the ground-truth. Plots of tremulous motion and estimation error are shown in Fig. 9.15.

Fig. 9.14. Plots of ground truth tremulous motion and the estimated tremulous position of the tool tip.

Fig. 9.15. Plots of tremulous motion and estimation error.

Table 9.1 Results of integration of sensing and filtering sub-systems in bench test

STD of tremulous motion (μm)	STD of Estimation error (μm)	Reduction (%)
18	5	72

9.5.2. Compensation performance of the instrument

The motion of the instrument tip is measured with M^2S^2 [Win 07; Latt 08a] for compensation performance evaluation. Plots of compensation performance of the bench test for x- and y- axes are shown in Fig. 9.16. A plot of tremor compensation results in absolute displacement is shown in Fig. 9.17. Power spectrum plots of the motion of the instrument tip in x-axis without and with compensation in a bench test are shown in Fig. 9.18 and Fig. 9.19, respectively. A plot in polar coordinates is shown in Fig. 9.20. Plots of compensation performance for the hand-held test are shown in Fig. 9.21. The words “OFF”, and “ON” in the figures respectively represent the period when the tremor compensation function of the instrument is turned off, and on. Power spectrum plots of the motion of the instrument tip in x-axis without and with compensation in a hand-held test are shown in Fig. 9.22 and Fig. 9.23, respectively. Table 9.2 shows compensation results of both the bench test and the hand-held test. Comparison of physical properties of Micron instrument and those of ITrem is shown in Table 9.3.

For the hand-held test, the instrument is held by a subject placing the instrument tip attached with a reflective ball within the workspace of M^2S^2 as shown in Fig. 9.24. The ball positions obtained from M^2S^2 are filtered using a 4th order band-pass filter with a pass-band range of 6-15 Hz to remove the intended motion, drift motion, and the measurement noise from M^2S^2 .

Fig. 9.16. Bench test results for x- and y- axes after passing through a high-pass filter.

Fig. 9.17. A plot of absolute displacement of tremor.

Fig. 9.18. Power spectrum of the motion of the tip in x-axis before compensation in a bench test

Fig. 9.19. Power spectrum of the motion of the tip in x-axis with compensation in a bench test

Fig. 9.20. A plot in polar coordinates (angles verses absolute displacements).

Fig. 9.21. Hand-held test results after passing through a high pass filter to remove drift.

Fig. 9.22. Power spectrum of the motion of the tip in x-axis without compensation in a hand-held test

Fig. 9.23. Power spectrum of the motion of the tip in x-axis with compensation in a hand-held test

Table 9.2 Results (Absolute rms amplitudes) of compensation performance in bench test and hand-held test

	OFF (μm)	ON (μm)	Reduction (%)
Bench test	26	9	65
Hand-held test	75	45	40

Table 9.3 Comparison of Micron and *ITrem*

	Micron	<i>ITrem</i>	Improvement (%)
Length (mm)	210	189	10
Weight (gram)	170	75	56

Fig. 9.24. (a) Setup for the hand-held test. (b) a close-up view of how the instrument is held for the hand-held test.

9.6. Discussion

In the bench test for the performance evaluation of sensing and filtering sub-systems integration, RMS value of tremulous motion is reduced from 18 μm to 5 μm by 72 % as shown in Table 9.1. In the bench test for the evaluation of tremor compensation performance of the instrument, when the tremor compensation function of the instrument is turned on, as can be seen in Table 9.2, RMS tremulous motion is reduced from 26.0 μm to 9 μm by 65 %. The result is near to the desired positioning accuracy of $\pm 10 \mu\text{m}$ for microsurgery [Charles 96].

In the hand-held test, the absolute RMS tremulous motion is reduced from 75 μm to 45 μm by 40 %. The compensation results for the hand-held test are not as good as those for the bench test. That might be due to several reasons such as the tremulous amplitude which is sometimes bigger than the manipulator’s workspace, and the possible inclusion of angular motion calculation errors.

Only the bench test results of tremor compensation performance of Micron are presented in [Ang 01; Riviere 03]. There is no report on performance evaluation of sensing and filtering sub-systems integration. Therefore, comparison between Micron and *ITrem* in the performance evaluation of sensing and filtering sub-systems integration cannot be performed. Since there is no report on the hand-held test, it seems that hand-held tests of Micron are not performed for some reasons, and hence comparison of hand-held tests results is not possible.

As can be seen in Table 9.3, the physical properties of *ITrem* such as size and weight have been improved comparing to those of the Micron. The maximum diameter of the instrument is 22 mm at the grasping section. Although it is still bigger and heavier than conventional surgical instruments, its length is now within the microscope constraint of 200 mm. The improvement is mainly due to the improvement in the sensing sub-system such as exclusion of a magnetometer [Latt 09b] which is employed in Micron, the placement of accelerometers strategically [Latt 08b; Latt 09a], and the integration design of sensing and manipulation sub-systems. The current compensation performance can be improved by addressing the following issues.

The values of displacement vectors among accelerometers are taken based on the dimensions of the mechanical drawing of the instrument. Although fabrication errors of the mechanical parts can be small, assembly errors and the errors which come from manual soldering of accelerometer IC chips on printed circuit boards (PCBs) can be significant. Estimation errors in tip motion due to these errors (inaccurate positioning of the accelerometers) can be significant depending on the magnitude of the angular acceleration and magnitude of these errors. For example, if the error in positioning of a particular accelerometer were ± 1 mm, then the worst case combined error for the two accelerometers would be 4 mm. The separation distance between the two accelerometers used for a particular angular acceleration is about 100 mm (please refer to Fig. 9.8). Therefore, there would be about 4% of angular acceleration error. As described in Chapter 5, multiplication of angular acceleration and the distance of about 90 mm (distance between the tip and accelerometer {2} in Fig. 9.8) would result in the tip acceleration error of about 360 mm/s^2 (720 mm/s^2 p-p). Translating from acceleration domain to position domain for 10 Hz motion, position error would be about $7 \text{ }\mu\text{m}$ p-p. In the current implementation of the instrument, these errors are not taken into account due to the lack of calibrating device. These errors can be collectively

known with calibration using a device or system which can provide 3 DOF angular motion of the instrument.

As can be seen from Fig. 9.8, the separation distance between accelerometers at {1} and {3} is 20 mm. The separation distance causes some errors arising from rotational motion of the instrument. The amount of error depends on the amount of orientation error of a_{3z} and magnitude of angular acceleration. Minimizing the separation distance will minimize the errors due to rotational motion.

Another possible source of error is accelerometer sensitivities which vary with orientation of the accelerometers. In the current implementation of the instrument, the effect of orientation on the sensitivities is not taken into account since the variation of the sensitivities is about 3% as shown in Fig. 9.25. Modeling of accelerometer sensitivities as a function of orientation of the instrument should be performed and included in the future implementation of the instrument. Block diagram of sensing sub-system to be implemented for further reduction of errors in Fig. 9.26.

Sensing sub-system's measurement noise which is the combination of accelerometer noise, noise coming from environment electromagnetic fields generated by nearby electrical equipments, noise in the data acquisition card, and so on is found to be about 10 mm/s² RMS. Plots of noise levels of acceleration measured by the sensing sub-system are shown in Appendix-B. Sensing and estimation quality of tremulous motion might be improved by addressing each of the noise sources. For example, noise due to electromagnetic fields can be reduced by carefully designing the electrical wiring layout so that the ground-loop is as small as possible. Shielding of the parts

Fig. 9.25. Variation of sensitivity (Scale Factor) of an accelerometer as a function of orientation [Ang 04].

Fig. 9.26. Block diagram of sensing sub-system to be implemented for further reduction of errors. Boxes and arrows drawn in dashed-line represent functions to be implemented in the future.

The manipulator having a larger work space covering all the tremulous amplitudes should be designed and used in the instrument. The manipulation sub-system should be evaluated with real tremor data rather than sinusoidal data and improvement should be made until its accuracy is within the acceptable limit.

9.7. Conclusion

A compact hand-held microsurgical instrument that performs active compensation of tremulous motion has been developed and is undergoing testing and refinement. The preliminary compensation results of *ITrem* are quite encouraging. The current results raise the hope for the success of the hand-held approach in microsurgery in the near future. The experiments performed so far are limited for a stationary task whereby it is attempted to position the instrument tip at a single point. In the future, path and trajectory dependent tasks will be included in the evaluation of the instrument performance.

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Chapter 10 : Conclusion and future work

10.1. Conclusion

In conclusion, all of the objectives set out on page 7 of section 1.3 are met. The performances of sensing and filtering sub-systems have been improved. The contributions that have been made by the author during the endeavor to meet the objectives are shown below in section 10.1.1. The publications that stem from the author's Ph.D. research are shown in appendix A.

10.1.1. *Contributions*

The contributions that stemmed from efforts to meet the objective 1) are as follows.

- Proposed a method of sensing tremulous motion of a surgical instrument tip using only accelerometers for real-time compensation. (Please refer to Chapter 4)
- Proposed placements of accelerometers for highest possible angular sensing resolution and introduced and implemented a compact design of the sensing sub-system of a hand-held active tremor compensation instrument. (Please refer to Chapter 5 and Chapter 6)
- Proved the assumption of very small angular velocities of instrument motion in micromanipulation tasks to ignore centripetal acceleration is correct. (Please refer to section 5.6)
- Proposed a method of elimination of phase and attenuation differences among accelerometer outputs for accurate angular sensing. (Please refer to section 6.3)
- Proposed a method of identification of accelerometer orientation errors and calibration for the errors. (Please refer to Chapter 7)

During an attempt to obtain high quality estimate of tremulous tool motion from the sensed motion for real-time compensation, which is objective 2), the following contribution is made.

- Introduced a WFLC- or BMFLC-based method of real-time estimation and prediction of tremulous tool motion from attenuated and phase-shifted sensed acceleration signals. Proposed the use of the method with analytical integration to eliminate position drift. (Please refer to Chapter 8)

The following contribution is a result of fulfilling objective 3).

- Design, implementation, calibration, and analysis of performance of an optical-based micro motion sensing system (M^2S^2). (Please refer to Chapter 3)

10.2. Future work

Erroneous motion of the hand contains non-tremulous erroneous motion: jerk and drift, and tremulous erroneous motion: physiological tremor. Riviere *et. al.* [Riviere 97a] reported that magnitudes of non-tremulous erroneous motion are often larger than those of physiological tremor. Jerk is a high-frequency motion which seldom happens. Drift is a low-frequency motion whose frequencies are less than 1Hz and overlap with the frequencies of intended motion. To compensate for the drift, low-frequency motion which is the combination of drift and intended motion has to be sensed and drift has to be distinguished from the intended motion. To distinguish drift from intended motion, research efforts are ongoing [Riviere 97b; Ang 01; Ang 02]. To sense the low-frequency motion, the accelerometers employed in the instrument are not suitable since the magnitudes of the drift and intended motion in acceleration domain are less than the accelerometer noise level. Therefore, as a future work, vision-based sensing of the motion is proposed to sense the low-frequency motion. Apart from compensating for the drift, the vision-based sensing will also facilitate some automatic functionality which will be described 10.2.1. Study of effect of instrument weight distribution and making the instrument diameter at the gripping area smaller is proposed in 10.2.2.

10.2.1. *Fusion of data from the hand-held instrument and vision*

Microsurgery and cell manipulations are usually performed under a microscope. To improve quality of tremulous motion sensing and facilitate compensation for the drift in these applications as well as to detect absolute tool tip motion for certain automatic functions which will be described later, a method is proposed in which a video camera is attached to the microscope, which will provide the same vision information as what the surgeon/operator sees. Since the camera has the same view as the surgeon/operator's, the tool tip motion and target position information is available from the camera all the time.

Example of an automatic function that can be performed is called “Snap-to-target” function which is a visuoservoing technique that is used to guide the tool to the target while the tremulous motion of the tool tip are being suppressed. When the human hand is used as a coarse manipulator to move the instrument close to the target(s), the system takes over to act as a fine manipulator to perform certain procedures that require no more than 3 DOF, e.g. to penetrate and inject drug into a blood vessel. This is illustrated in Fig. 10.1.

Fig. 10.1. (a)The instrument deflects its own tip to cancel the tremulous motion of the tool tip; (b) On top of compensating for tremulous motion, the vision system determines the relative position of the target to the tool tip and ‘snaps’ the tip to the target.

Tool tip position information obtained from accelerometers will be fused with that obtained from vision. It should be noted that the tip position information from accelerometers is the complete 3 DOF position vector while that from vision is only 2 DOF if only one camera is employed.

The block diagram of fusion of IMU data and vision data is shown in Fig. 10.2.

Fig. 10.2. Block diagram of fusion of data from the hand-held instrument and vision.

Fig. 10.3. Illustration of camera reference frame $\{C\}$ and body frame $\{B\}$.

Fig. 10.4. Block diagram of proposed implementation of fusion.

Block diagram of proposed detail implementation of fusion is shown in Fig. 10.4. Tremulous position of the tool tip obtained from accelerometers in the instrument is with respect to the steady tip frame $\{S\}$ while that from the camera is with respect to the camera frame $\{C\}$. In order to fuse the two pieces of information, they must be in the same coordinate system. Tremulous position from the instrument is transformed into camera coordinates first. Then fusion is performed using fusion gains. Then, the fused information in camera frame is converted back to the steady tip frame for compensation. For coordinate transformation, a rotation matrix which relates the camera frame to the steady tip frame is required.

The symbol ${}^S R_C$ represents the directional cosine matrix (DCM) which relates camera frame to steady tip frame. The symbol ${}^C R_S$ which is the transpose of ${}^S R_C$ is the directional cosine matrix which relates steady tip frame to camera frame. Tilt and roll angles of the instrument will be obtained from accelerometers. Pan angle information will be obtained from vision. Using these angle values, DCM can be calculated. The

symbols k_i and k_c are fusion gains which can be obtained from error variance of tip position calculated by the instrument and that by camera.

10.2.2. Study of effect of instrument weight distribution and reducing gripping diameter

Weight distribution of the instrument might affect tremor amplitude. Well-balanced instruments might prevent the amplitude of the tremor from becoming large. The effect of weight distribution should be studied by placing an additional piece of dense material at different locations on the instrument along the instrument axis.

Currently, the diameter of the instrument at the gripping area is 22 mm. The size of the diameter is limited by the presence of actuators inside the manipulator. Generally, the gripping diameter should be about 10 mm [Patkin 77] for better ergonomics. To reduce the diameter at the gripping area, re-designing of the manipulator sub-system which is currently limiting the size is one of the options. Another option is to re-design the instrument so that the gripping area is just before the manipulation sub-system.

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Appendix A

Publications stemming from the author's PhD research, contributed as a first author, are listed below.

Journal:

1. **W. T. Latt**, U-X. Tan, C. Y. Shee, C. N. Riviere, and W. T. Ang, "Compact Sensing Design of a Hand-held Active Tremor Compensation Instrument," *IEEE sensors journal*, vol. 9, no. 12, pp. 1864-1871, Dec., 2009.
2. **W. T. Latt**, U-X. Tan, C. Y. Shee, and W. T. Ang, "Design, implementation, and calibration of an optical micro motion sensing system," *IEEE sensors journal*. (Under Review)
3. **W. T. Latt**, U-X. Tan, K. C. Veluvolu, and W. T. Ang, "Drift-free Position Estimation of Periodic Motion using Inertial Sensors," *Sensors and Actuators A: Physical* (Under Review).
4. **W. T. Latt**, U-X. Tan, C. N. Riviere, and W. T. Ang, "Placement of Accelerometers for High Sensing Resolution in Micromanipulation," *Sensors and Actuators A: Physical* (To be submitted).
5. **W. T. Latt**, U-X. Tan, K. C. Veluvolu, and W. T. Ang, "Real-time estimation and prediction of physiological tremor," *IEEE Trans. Biomed. Eng.*, (To be submitted).

Conference:

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Appendix B

Fig. B.1. Plots of noise levels of acceleration measured by the sensing sub-system.