INTRODUCTION

The World Health Organization (WHO) has reported that cataracts are responsible for 51% of the total cases of world blindness. An estimated 20 million people in the United States alone currently suffer from cataract-induced vision impairment [1]. The WHO has also found that coronary heart disease was the cause for approximately 7.4 million deaths in 2012, representing approximately 13% of total mortalities [2]. Though effective, current surgical techniques for treating cataracts and CHD rely heavily on the skill and experience of surgeons. During microsurgical procedures, surgeons use visual feedback when forces are below the threshold of touch [3]. Percutaneous coronary intervention, or angioplasty, similarly requires surgeons to focus on an external monitor displaying an angiogram while concurrently performing the arterial procedure [2]. In effect, both microsurgery and angioplasty techniques require a surgeon’s precise and steady control over visually indistinguishable and tactilely insensitive membranes, nerves, and blood vessels under a microscope [3]. Even with the use of a microscope, surgeons cannot tactiley assess their movements, creating the potential to accidentally rupture blood vessels and membranes.

Simulation devices, or phantoms, have become important tools for microsurgeons to practice tissue puncture experiments [3]. Phantoms are devices capable of rendering physical surfaces by using force feedback to create surface characteristics, such as texture, viscosity, and stress and strain. The Geomagic Touch and Magnetic Levitation Haptic Device (MLHD) are two such haptic platforms that are commercially available for surface rendering. The MLHD can move in six degrees of freedom and uses a flotor surrounded by a bed of magnets to generate forces onto the user’s hand [3]. However, the MLHD has been shown to be ineffective at rendering biological membranes due to limitations in force output and inertial effects caused by its heavy flotor [3]. The 1 Degree of Freedom Haptic Renderer (1DOF) was developed to address the limitations of existing haptic platforms. It uses a woofer loudspeaker actuator with force and displacement sensors to simulate tissue interaction [3].

OBJECTIVE

A Proportional-Integral-Derivative (PID) controller is to be designed and coded to model two spring simulations to test the viability of the 1DOF as a sensitive membrane modeling device.

SUCCESS CRITERIA

Two criteria must be satisfied to confirm the viability of the 1DOF. The first is to develop a zero-stiffness (ZS) spring model, which responds to low forces and acts as a spring with a small spring constant model must allow the 1DOF to move freely over its displacement range with little resistance. The second criterion is to develop a virtual wall (VW) spring that outputs current to absolutely resist movement. It should simulate a spring with a large spring constant and not displace with any amount of force.

METHODS

The 1DOF actuator is an 80W subwoofer speaker (Faital PRO 5FE120) with a 5 inch (12.7 cm) diameter and a cone mass of 11 grams. It is capable of 9.5 mm displacement from rest and can move steadily against forces up to 10 N [3]. Speakers have the advantage of direct electromechanical transduction, with force related simply to the voltage across the speaker coil. For sensing purposes, a stereolithographic scaffold with a Honeywell FS03 Force Sensor was placed on top of the speaker cone. An optical IR transceiver (Vishay TCRT5000L) was mounted and used to measure speaker cone position. The total resulting mass of the cone including the sensor was 23.6 grams. The renderer is controlled by an Analog Devices ADuC7026 microprocessor. A Wixel USB module (Pololu; Las Vegas, NV) enabled communication between the “master” microprocessor and a “slave” computer for data logging and mode selection.

A Proportional-Integral-Derivative (PID) feedback controller was used to regulate speaker current based on force and displacement measurements [4]. The error (e) was determined by subtracting the sensor voltage at some set point from the current sensor voltage. Proportional control multiplies a proportional gain (Kp) by the error, while the integral (Ki) and differential (Kd) controls summate previous errors and predict future errors with their respective gains [4]. The two primary modes of PID
were developed: (1) Virtual Wall mode, in which a set-point was chosen based on the desired displacement, with the error used to resist displacement from the set-point with maximum force, and (2) Zero Stiffness mode in which speaker voltage was used to minimize force detected by the sensor and thereby simulate open air. In each case, PID control was used determine the output voltage based on the error as shown in Eq. 1.

\[ V_{out} = K_p \cdot e(t) + K_i \int_0^t e(t') \, dt' + K_d \frac{de(t)}{dt} \]

Eq. 1: PID controller for output voltage

Open-loop Ziegler-Nichols tuning was performed to empirically determine the PID gains from the ultimate gain \( K_u \), the value of \( K_p \) at which the system began oscillating, and the period \( T_u \) of oscillation for each mode. Anti-windup control was implemented by scaling down the integral component by the difference in the calculated controller output and the maximum actuator output [4]. A slew-rate limiter and a low pass filter were added to compensate for high frequency signal noise accentuated by the derivative gain term.

RESULTS

Tyreus-Luyben PID controller settings were adopted over Ziegler-Nichols due to their greater stability with the system. Final calculated \( K_u \) and \( T_u \) are shown in Table 1. PID control of Virtual Wall mode produced forces opposite to the user, resulting in almost zero displacement as seen in Figure 1. Similarly, the new rendition of Zero-Stiffness mode actively removed the tension at all positions of the speaker cone as seen in Figure 2.

<table>
<thead>
<tr>
<th></th>
<th>Virtual Wall</th>
<th>Zero-Stiffness</th>
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<tbody>
<tr>
<td>( K_u )</td>
<td>1.50</td>
<td>7.348</td>
</tr>
<tr>
<td>( T_u )</td>
<td>0.025</td>
<td>0.002</td>
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Table 1: Calculated \( K_u \) and \( T_u \) Values

Figure 2: Actuator in virtual wall mode receiving large push/pull forces, but undergoing almost zero displacement (red, middle)

Figure 3: Actuator in zero stiffness mode undergoing rapid changes in displacement and detecting no forces (blue, top)

DISCUSSION

Dynamic control of the actuator was achieved with PID and successfully rendered both a very stiff membrane in VW mode and an empty space simulation in ZS mode. In both modes, PID control of the speaker actuator minimized errors with lower latency than simply P or PI methods. The addition of derivative control predicted future errors and balanced the error averaging conducted by the integral term. Prior to anti-windup control, the actuator suffered from repeating integration of previous errors. Anti-windup eliminated “stickiness” related to actuator saturation at the ends of the displacement range. Implementation of a low-pass filter resulted in a decrease in sensor noise interference and windup due to large derivative terms from higher frequency inputs.

REFERENCES


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