DESIGN OF A FLEXIBLE OPTICAL TRACKER FOR ORTHOPAEDIC NAVIGATION

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INTRODUCTION

Optical tracking has become the standard for measurement in computer assisted orthopaedic surgery (CAOS). However, the general bulkiness of the rigid tracker components hinders the usability of navigation technology. The trackers must be designed sufficiently large to separate the markers for measurement resolution, and stiff enough to ensure virtually no mechanical deformation to maintain the rigid-transformation between the markers and the body for risk of losing registration. The trackers' bulkiness serves as a source of frustration for orthopaedic surgeons.

The most severe systematic error for CAOS systems is a displacement of the tracker with respect to the bone (Leardini 2013). There have been reports of post-operative stress fractures and soft tissue damage resulting from the use of bicortical tracker fixation pins (Lionberger 2007; Ossendorf 2006). The size of conventional tracker arrays has meant that navigation technologies are generally not well suited for applications with smaller bones, such as the patella or scaphoid (Smith 2014). An ideal tracker would be accurate, small, non-obtrusive, and widely applicable (Welch 2002). Current tracker designs fail on several of these counts.

The purpose of this paper is to present a new design for CAOS trackers to overcome these limitations, and to demonstrate the concept's feasibility through simulations and experiments inspired by a total knee arthroplasty application.

DESIGN, THEORY, AND IMPLEMENTATION

TRACKER DESIGN

Unlike conventional CAOS trackers, which feature a cluster of markers on a single rigid frame (Figure 1a), our new design positions the markers on the ends of separate low profile flexible pins (Figure 1b, c). This design permits for the markers to be physically deflected without significantly affecting the overall tracking accuracy. The tracker's modularity allows for a longer baseline distance between measurements (conceivably as long as the bone) which leads to improved angular resolution and accuracy. The contact forces (applied moment) at the bone-pin interface resulting from accidental bump would be less than for a rigid tracker because of the flexibility and small size, which could enable stable unicortical tracker fixation.

CONSTRAINT THEORY AND PIN FLEXION

We cannot rely on conventional assumptions of tracker rigidity because the markers are able to move relative to one another. However we can still constrain the body by exploiting each pin's stiff axial constraint, and ignoring transverse deviations of the markers. Using this approach, we require six flexible pins (or three pins with twinned markers) to fully constrain the bone. We used Euler-Bernoulli beam theory to model the flexion of our pins as slender cantilever beams, with one end anchored into the bone and the marker mounted to the free end. As illustrated in Figure 3, the trajectory of the marker with respect to the anchor may vary with different loading conditions, which introduces a potential source of error. To minimize the effect of such variation, we assume a nominal bending model based on a point-load applied at the middle of the pin.

IMPLEMENTATION ALGORITHM

We used an unscented Kalman filter (UKF) algorithm to implement our model for the flexible tracker, and enable efficient real-time operation. The state vector x describes the bone's pose (position and orientation), and the deflection of each of the flexible pins, where $(x \;\; y \;\; z)$ is the bone position, $(\phi \;\; \theta \;\; \psi)$ is the orientation, and $(\overrightarrow{P_1} \; \cdots \; \overrightarrow{P_N})$ is the deflection state of each pin.

$$x_k = [\, x \;\; y \;\; z \;\; \phi \;\; \theta \;\; \psi \;\; \overrightarrow{P_1} \; \cdots \; \overrightarrow{P_N} \,]^T$$

The process model $F(x)$ is a function describing the anticipated state progression over time. In the absence of any external information on the evolution of the system's state, we describe the process as a constant position, white noise velocity model.

$$\hat{x}_{k+1} = F(x_k, w_k) = x_k + w_k$$

The measurement model $H(x)$ is a function linking the state space to the measurement space.

$$\hat{y} = H(x) = T_G^B \times T_B^A \times T_A^P \times f(T_A^P) \times \vec{d}_P^M$$

The measurement model describes a series of Euclidean rigid-body transformations between the sensor-coordinates and the marker points, as shown in Figure 2. T_G^B is the transformation between global and bone coordinates, which is the end goal for our navigation system. T_B^A is the unchanging transformation between body and anchor coordinates, which can be acquired with a tracked pin insertion tool or a digitization step prior to use. T_A^P is the transformation from anchor to pin coordinates, which describes the deflection of the pin. The function $f(T_A^P)$ accounts for pin bending nonlinearity, which is a function of the deflection magnitude specific to the expected model of pin bending. Lastly, the vector d_P^M describes the geometry of the pin in local coordinates.

METHODS

We conducted a series of simulations and experiments of our flexible tracker system, inspired by a total knee arthroplasty application. The system was tested under different operating conditions to establish the system's baseline tracking performance.

EVALUATION METRIC

We combined translation and rotation error components into a single combined evaluation metric (CEM). The rotation component is scaled by a characteristic length factor to represent a projected targeting error, for which we used half of a typical transepicondylar distance (~50mm).

$$CEM = \sqrt{\Delta X^2 + \Delta Y^2 + \Delta Z^2 + (\, f \, \Delta\theta \,)^2}$$

In simulations, our estimated pose components were compared against the ground truth (for accuracy). In static experiments, the estimated pose was compared against the trial's mean pose to calculate a combined precision metric, and in one case we compared our pose estimation to the pose of a rigid tracker for a combined discrepancy metric.

SIMULATION METHODS

Our simulations of the flexible tracker system were run using MATLAB 2013. We simulated marker jitter which we established experimentally by repeatedly sampling a stationary point. (0.006mm $\vec{x}$, 0.006mm $\vec{y}$, and 0.025mm $\vec{z}$). The trials ran for 300 frames at 30Hz (ie. 10s). We conducted simulations in which: the body was kept motionless and all geometrical parameters were assumed known without error; the body was moved at a constant rate for a range of speeds in both translation (0 to 3000mm/s) and rotation (from 0 to 112.5 º/s); the body was kept static while a pin was deflected by varying magnitudes (0 to 40mm lateral deflection); and with typical uncertainties: we tested bending from different loading conditions, and simulated error in anchor digitization.

EXPERIMENT METHODS

We conducted experiments to validate our simulation results. Using a set of prototype flexible tracker pins, a Northern Digital Inc. Polaris Spectra optical sensor, and a piece of wood to represent a femur, we recreated a surgical navigation system. We conducted experiments in which: the body was kept motionless; the body was manually moved throughout the measurement volume; the body was kept static while pins were manually deflected; and the body was attached to a high-precision linear translation stage for a comparison with a gold-standard measurement.

RESULTS

Graphical summaries of the simulation and experiment results are shown in Figure 4 and Figure 5 respectively.

SIMULATION RESULTS

In our static simulations the median combined tracking error was 0.009mm (90% CI: 0.004 – 0.017mm). The system was insensitive to translations up to speeds of 3000mm/s. In rotation, the tracking error increased proportionately with motion speed, with a median error of 0.40mm (0.39 – 0.42mm) at 112.5º/s rotation. The rotation error disappeared after the motion ceased. The system's sensitivity to bending is affected by the particular loading conditions. Our tracker is insensitive to moderate bending conditions. However, for more extreme bending conditions, we found tracking errors of 0.70mm (0.68 – 0.71mm) for lateral deflections of 30mm.

EXPERIMENT RESULTS

The static body was tracked with a median combined precision of 0.03mm (0.01 – 0.07mm). During freehand motion, the system had a median discrepancy between the rigid and flexible tracker estimates of 0.19mm (0.10 – 0.25mm) in translation, and 0.19mm (0.04 – 0.75mm) in Rotation. In flexion, the approximated mid-pin point load of 50mm lateral deflection resulted in a pose deviation of 0.48mm. In the pin-tip loading case, the deviation was 0.84mm at 50mm, which exemplifies the errors resulting from variability in loading conditions. The constrained translation case showed a mean accuracy of 0.12mm (0.04 – 0.23mm) for cross-plane translations, and 0.04mm (0.03 – 0.05mm) for in-plane translations.

DISCUSSION

The flexible tracker system provided accurate and precise estimations of the bone's pose in typical operating conditions, and it demonstrated robustness to marker deflections. The precision measured in our static experiment is comparable to that of a conventional rigid tracker, and our findings are consistent with results from Khadem (2000) who reported rigid-tracker static precision of 0.07mm RMS using the NDI Polaris sensor. The flexible tracker demonstrated an insensitivity to translation motion with no marked increase in error compared with the static case. In rotation, however, tracking error increased proportionally with speed, which is an artefact of the UKF algorithm handling rotations, which are nonlinear transformations. These rotation errors remain modest (<0.5mm for 90º/s), and disappear as the rotation stops. Furthermore, surgical actions will not occur during motion, though this rotation error may restrict the tracker from use in more dynamic applications. The results of our constrained translation experiments are consistent with a study by Wiles (2004), who reported translational accuracy of 0.185mm. The studies by Khadem (2000) and Wiles (2004) were conducted prior to the launch of the Polaris Spectra sensor (2006), which has an improved measurement accuracy by 0.1mm RMS. This partly explains the discrepancy in our respective findings.

Tracking errors from pin flexion increase proportionally with the magnitude of the deflection. The severity of these errors is further affected by the closeness between the applied pin bend and our implicit bending model. The tracking errors were marginal for considerable bends, less than 0.5mm for 40mm lateral deflections.

CONCLUSION

The flexible optical tracker is intended to serve as a drop-in replacement for conventional rigid tracker components. By designing a small, modular and flexible tracker system, we aim to overcome many of the shortcomings of rigid trackers, thereby improving the user experience for orthopaedic navigation.

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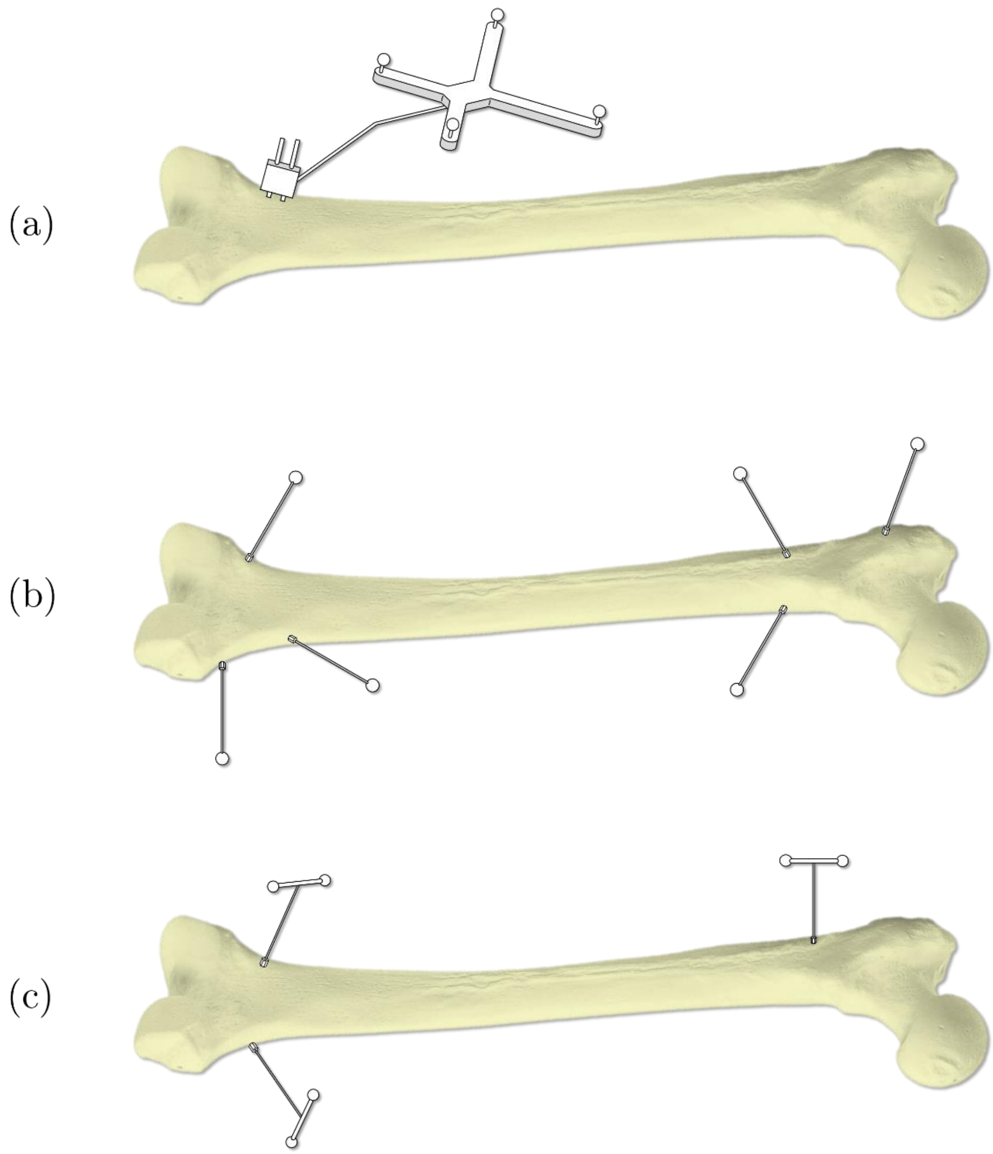

Figure 1: A comparison between optical tracker designs, showing: (a) conventional rigid tracker, (b) single-marker flexible tracker, (c) double-marker flexible tracker.

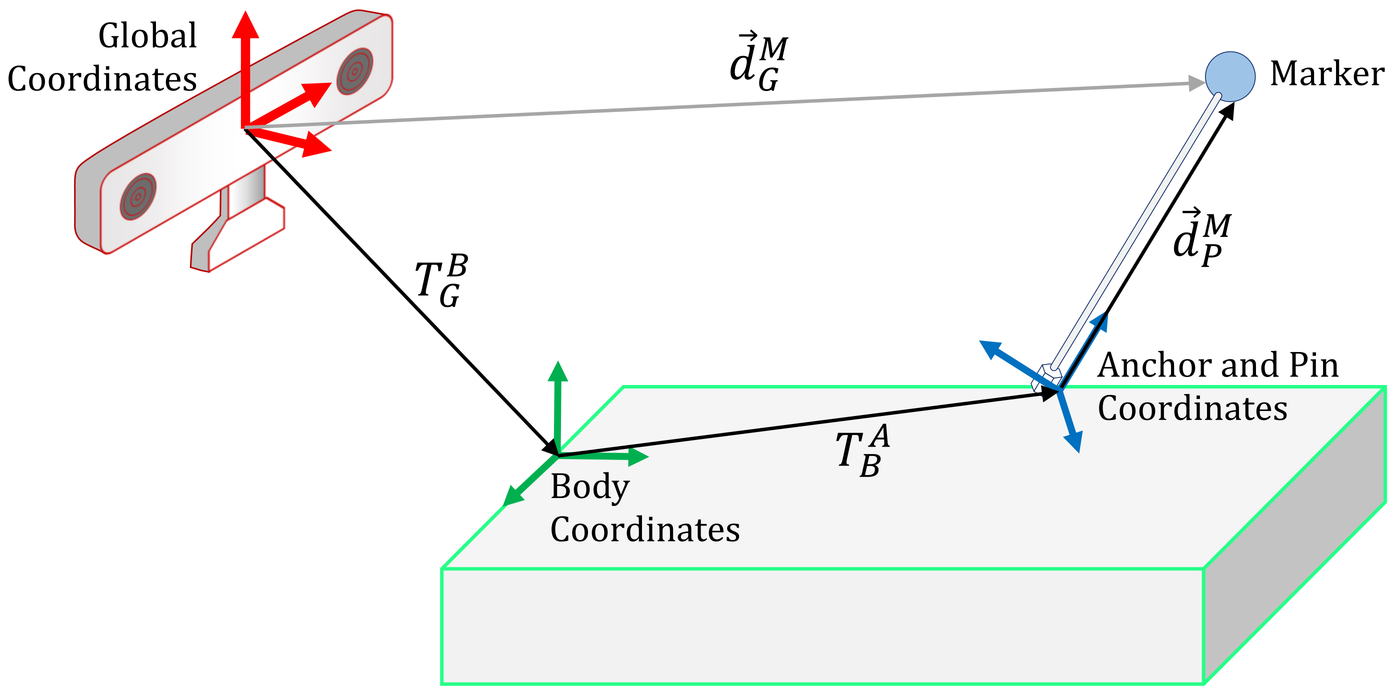

Figure 2: Geometrical formulation of system, showing the kinematic chain of rigid-body transformation between the position sensor and one pin. The transformation T_G^B is the end-goal for our system, as it represents the location and orientation of the underlying bone segment.

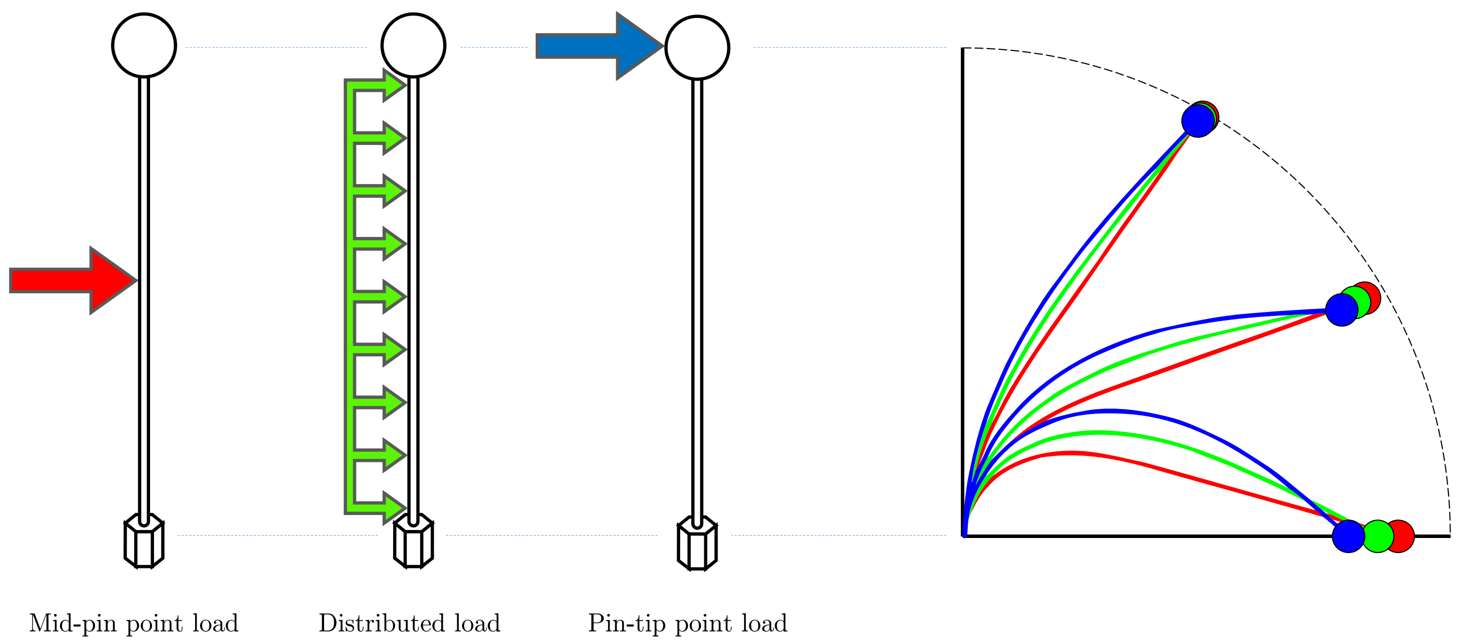

Figure 3: A comparison between three loading cases, demonstrating the variability in bending. For example, the pin-tip load induces a greater contraction between marker and anchor than the mid-pin load. We used a mid-pin load as our implicit bending model for this study.

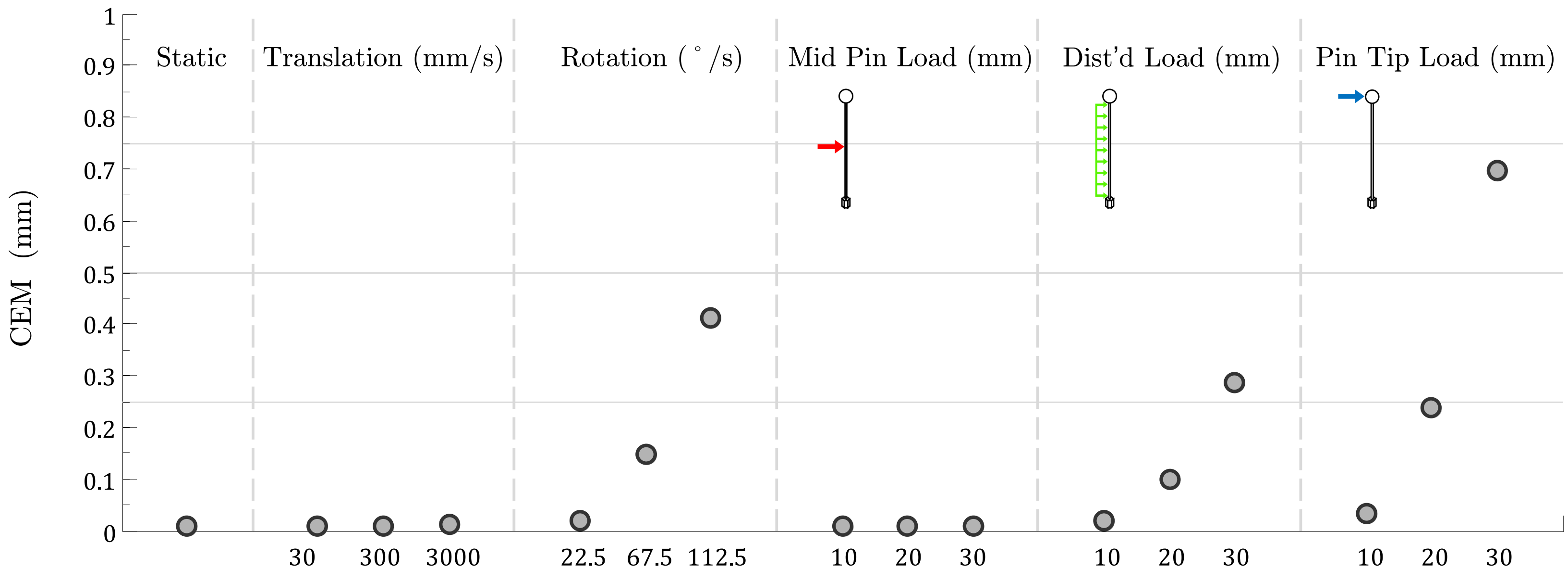

Figure 4: A summary of simulation results. Accuracy was evaluated as a combined metric comparing the pose estimation with the ground-truth. Error increases proportionally with rotation, but remains less than 0.5mm for rotational speed of 112°/s, which is greater than we would expect to see in the operating room. Error from pin bending varies with loading conditions, and is proportional to the magnitude of the deflection.

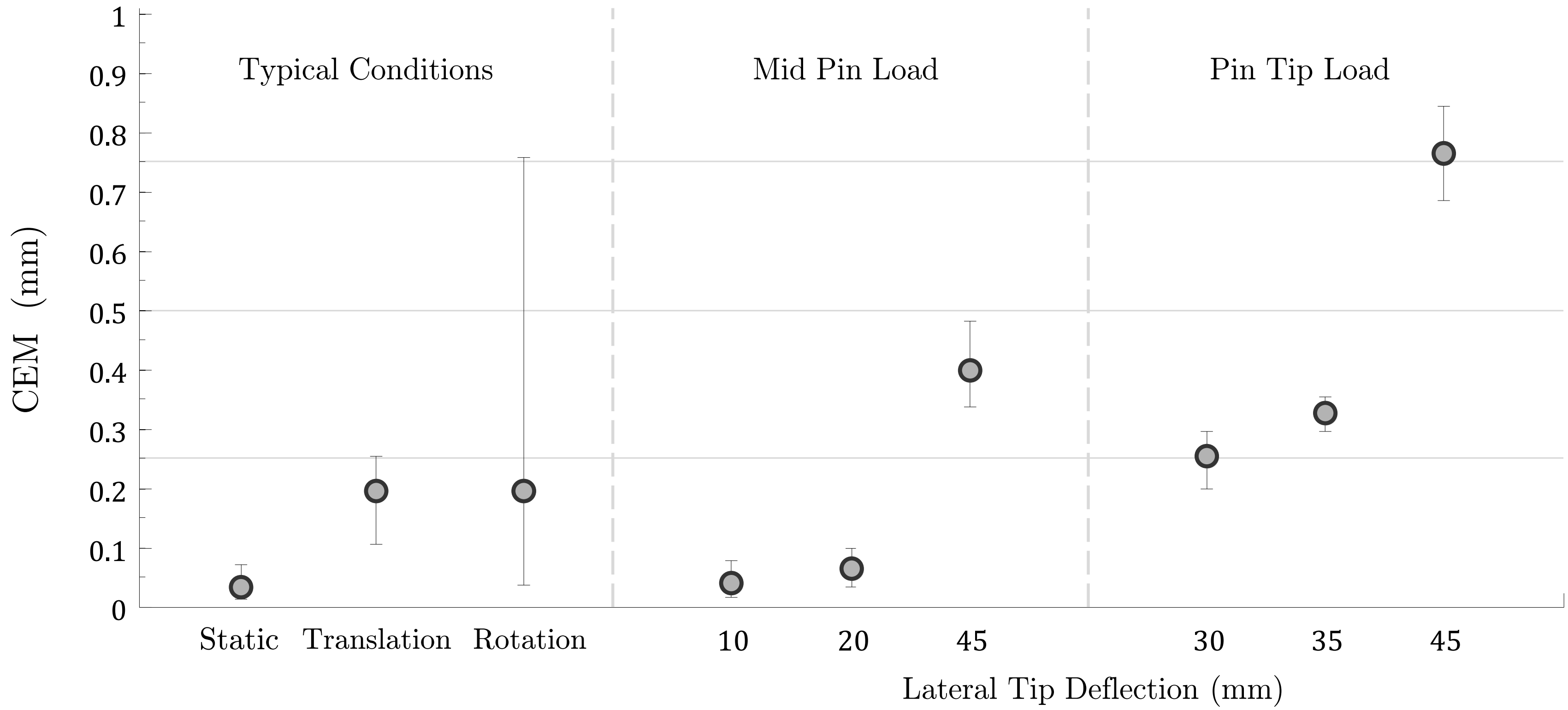

Figure 5: A summary of experiment results. The translation and rotation results are reported the discrepancy between the estimated pose and measurements made by a rigid tracker. The static and bending results are tracking precisions. The tracking error is below 0.5mm for all cases except a 45mm pin tip deflection, and the upper bound of the freehand translation.