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Trajectory Control for 3 Degree-of-Freedom Wrist Prosthesis in Virtual Reality: A Pilot Study

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***Abstract*— Controlling a complex upper limb prosthesis, akin to a healthy arm, is still an open challenge due to the inadequate number of inputs available to amputees. Designs have therefore largely focused on a limited number of controllable degrees of freedom, developing a complex hand and grasp functionality rather than the wrist. We introduce a novel 3 degree of freedom wrist trajectory control which takes advantage of joint coordination that aims to vastly simplify its use in a prosthetic device. We demonstrate its efficacy through a series of tasks performed by participants in a virtual environment. Trajectory control enables users to complete tasks faster with a more intuitive interface without additional body compensation, while featuring better cosmesis when compared to sequential and simultaneous control.**

I. Introduction

In prosthetics, grasping has overshadowed the role of orientating the hand itself in enabling us to complete basic activities of daily living (ADL) [1]. A lack of a wrist leads to unnatural movements [2] and amputees developing overuse syndromes [3]. The absence of an intuitive control interface is part of the reason why 3 degree of freedom (DOF) wrist devices are not mainstream on the market, which does nothing to abate the high prosthesis abandonment rates [4]. In this paper we propose and demonstrate in virtual reality (VR) a novel control approach (Fig. 1) that when implemented would make use of all 3 DOF of a myoelectric prosthetic wrist while vastly simplifying the cognitive burden of control using only 2 inputs; easily applicable to standard 2-site surface electromyography (sEMG) control interface. The controls consist of interpreting a single DOF input into predefined multi DOF trajectory modes, five in total, discovered in the authors' previous work on arm motions performing ADL [5]. These modes are based on the idea of joint angle coordination [6], obtained by recording, clustering and averaging arm motions from healthy participants performing an extensive set of ADL tasks. We coin the proposed approach "trajectory control", and we believe would offer users more intuitive, better motion cosmesis, and more assistive control while limiting body compensation. We test this hypothesis using various quantitative and qualitative evaluation metrics [7] that pin the proposed trajectory control against state of the art myoelectric control methods.

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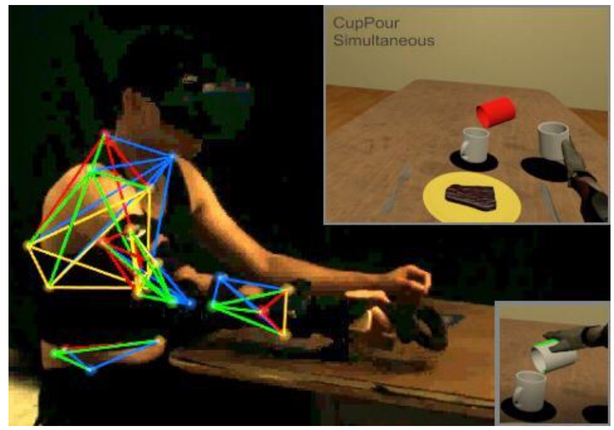


Figure 1. Subject performing a cup pouring task, seen wearing the HMD and is using the controller to operate the virtual wrist. In the top right a semi-transparent red cup is visible indicating the desired cup orientation goal, which turns green (bottom right) once the cup reaches the target.

Advancements in prosthetic wrist devices [8] are inadequate without an intuitive or practical way for amputees to control them [9], and various groups have attempted to bridge the gap. One approach is to determine synergies between the residual limb, shoulder or elbow angles, and the wrist [10]–[12]. These methods would make use of sEMG signals or kinematic data obtained from inertial measurement units (IMUs) from the residual limb, and are trained, using artificial neural networks, to interpret user intention and activate the proper control response. Although these are robust under certain conditions, aside from requiring additional sensing equipment, they are impractical for mass use due to the very involved training and retraining phase that is unlikely to reach the majority of amputees. Wrist rotation could instead also be directly coupled to shoulder abduction, which has the potential to reduce compensatory movement [13]. Bimanual manipulations have also inspired a wrist device control that mirrors the opposing healthy wrist by taking advantage of symmetric or anti-symmetric motions [14]. These proposed approaches offer users an additional control input beyond the two-site sEMG input [15], and while they enable users to perform complex wrist motions, they also impose a cognitive burden that limits efficacy. The proposed trajectory control, on the other hand, does not require additional control input and could easily make use of an existing sEMG prosthesis socket.

Virtual reality (VR) and has been embraced across wide range of applications [16], including within the prosthesis community, as a cheap way to iterate and test new device designs and controls that extend to a real world setting [17], [18]. Other applications include training to use a prosthesis [19], fitting [20], and rehabilitation [21], [22]. Virtual reality environment (VRE) has also been demonstrated to be convincing enough to treat phantom limb pain [23], [24], further reassuring its use to test our proposed controls. In this work we make use of VR (Fig. 1) to test the novel 3 DOF wrist control on able-bodied subjects, while avoiding having to build a wrist device itself. By exhibiting controls' efficacy, we hope to provide a foundation for the development of a physical 3 DOF wrist prosthesis.

This paper builds upon the authors' previous work [5], [25] by looking to demonstrate the angle trajectories as feasible control modes, particularly for implementation in a 3 DOF wrist device. For full description of how the trajectories were obtained we direct readers to [5]. An additional summary Is included below in section II. D. In the present work we aim to address the following questions regarding their use in a prosthetic device: do trajectory modes reduce the time it takes to align the hand, do they mitigate body compensation, and do users have a preference. For the remainder of this paper, we begin with describing the experiment set up, protocol, and analysis tools (section II). We proceed with the results (section III) and follow with a discussion and propose future work (section IV).

II. Methods

Throughout the experiment, participants were asked to perform ADL tasks while controlling a virtual prosthetic device. A novel VR interface is created using [26] as a blueprint (Fig. 2). Two participants took part in this initial run of the study (Table I). Additional participants are planned for future work. These were healthy right handed individuals without motion or visual impairments, and were comfortable being immersed in VR. Data collected from the experiment was exported and analyzed using MATLAB 2019a. This study protocol was approved by Yale University Institutional Review Board, HSC# 1610018511.

TABLE I. Participant Characteristics

Participant	Sex	Age	Weight (lbs)	Height (inches)	Arm Length (inches)[1]
P1	M	27	135	67.5	27.5
P2	M	23	160	70	28

[1]Measured from the shoulder to the tip of the middle finger

A. Virtual Reality Set Up

The study was designed to provide participants a fully immersive VRE, where they can view a virtual arm in place of their own and complete ADL tasks as they would in real life. We used Unity platform to create the VRE as well as to interface with the inputs streaming from the users. Virtual objects were either designed within the Unity environment, created in Maya 3D modeling software, or obtained for free from the web. Object models were scaled to the task requirements, while the virtual prosthetic hand, forearm, and humerus models were scaled to the average human arm dimensions. Since the focus of testing wrist control is to correctly orient the hand, rather than grasping, we turned off kinematic and dynamic model interactions, and omitted the need for the participants to grasp the objects. Therefore, after the reaching task, objects were automatically placed within the hand akin to how they would be grasped when using a natural hand.

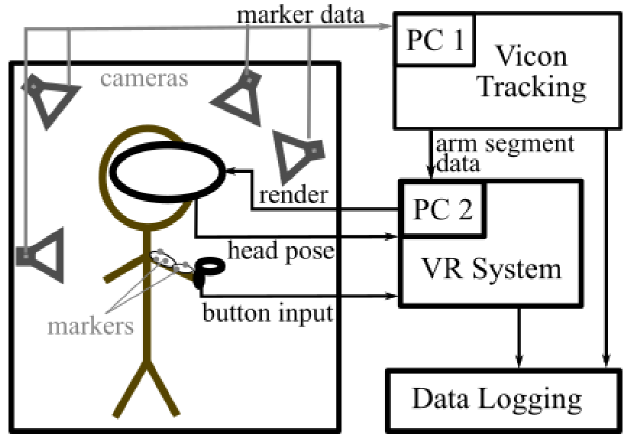


Figure 2. Flowchart of the experiment set up. Data from the user is collected through the motion capture and transmitted to PC 1 where arm segments are reconstructed. Segment data is sent to PC 2 where the virtual arm is matched to mimic the location and orientation. User receives visual feedback through the HMD and moves the wrist using the Vive controller held in the same hand.

B. Control Inputs

Motion tracking has been shown to be an effective input to kinematically controlling a hand in VR [27]. In this study we used 12 Vicon Bonita cameras to track the subjects' hand and forearm, in the case of natural motion control, and forearm and humerus, for all other conditions. In addition to providing participants an immersive experience by displaying a virtual arm in place of their own, this also offered them a reference when operating the simulated prosthetic device. The respective reflective marker clusters are shown in Fig. 3. Additional markers that were placed around the subject's humerus, torso, and pelvis were not streamed into Unity, but were included for body compensation analysis. The reflective marker locations include bony landmarks (seen in Fig. 1) that were used to calculate the joint coordinate reference frames according to [28].

Participants used a Vive controller, rather than an sEMG input, to operate their simulated prosthetic devices, providing the benefit of an intuitive interface that required virtually no training. This streamlined the training process for able bodied participants who might not be as familiar with an sEMG interface, thereby emulating experienced powered device users. Specifically, the standard contralateral two-site sEMG control is imitated by using two antagonistic buttons on the Vive controller, one near the index and one near the thumb. If trajectory control outperforms other types of control using the Vive controller, then we claim that this difference in control performance would appear when using sEMG as well. We have briefly tested this claim [29], but hope to expand on this in future efforts. Head tracking was also performed through HTC's head mounted display (HMD). Calibration between the two systems, Unity and Vicon tracking, was performed by matching the Vive controller as it was tracked by the HTC cameras and Vicon.

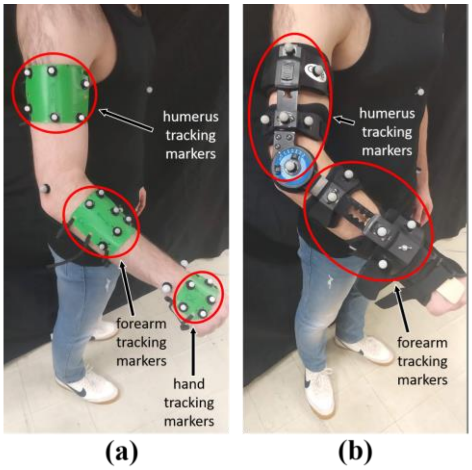


Figure 3. (a) Marker arrangement for the positive control; participants' arms were unconstrained. Hand markers were used to control the location and orientation of the virtual hand while the rest of the markers were exported for further body motion analysis. (b) Braced condition used for the other four control modes. Forearm markers were used to control the virtual forearm, while the virtual hand was either fixed in place (negative control), or operated using the hand held Vive controller. Wooden piece was inserted into the brace to ensure a fixed flexion-extension position. The elbow brace hinge was given full range of motion.

Figure 4. Standard HTC Vive Controller. Vicon reflective markers were placed in a known arrangement around the controller's head and were used to calibrate the virtual space between Vicon and HTC.

C. *Control Modes*

Participants were asked to perform each set of ADL tasks using five control modes, identified as follows: positive, negative, sequential, simultaneous, and trajectory control. Positive control simply refers to the use of an unencumbered hand, and serves as a performance target that prosthesis users aspire to achieve. In this mode participants saw the virtual prosthetic hand closely match the position and orientation of their real hand. We also refer to this mode as natural control.

For the other four modes, the wrist was fixed in place using a custom made wrist brace (Fig. 3b), similar to the one used in [2]. The orthopedic wrist brace (DonJoy ComfrotFORM Wrist Support Brace – DJO Global, Vista CA, USA) was combined with an elbow brace (Orthomen ROM Elbow Brace) using Velcro straps and a bolt. The elbow brace was given full range of motion and served to limit the pronation-supination of the wrist without hindering elbow motion.

A negative control was included in the experiment, a condition in which the virtual prosthesis lacked wrist mobility. For this mode, subjects had to complete tasks without the ability to rotate their wrist and had to compensate for it using their residual limb and torso. Although task performance was closely monitored by the experimenters, subjects were additionally instructed to indicate if a task could not be completed.

Sequential control aimed to represent current myoelectric technologies on the market where users have access to only two control inputs. The Vive controller's (Fig. 4) trigger and trackpad button took the place of the two-site contralateral sEMG that are often placed on amputees' forearm; in the case of powered prosthetic devices. On the controller, the trigger drove the wrist forward along a specified direction and the trackpad button drove it backward. A simultaneous press of both inputs switched the DOF along which the wrist rotated. When switching between DOF, the order of rotations cycled from pronation-supination, flexion-extension, to radial-ulnar deviation. This mode switching scheme is often used to switch between grasps in powered prostheses.

Simultaneous control granted individuals access to all control inputs at the same time by leveraging all the buttons that were available on the Vive controller. Similar to the positive control, this mode represented the state of the art and a theoretical condition where users have 6 control inputs available to them. The Vive controller's trackpad is a single button with location sensing in two dimensions, which we took advantage of by offering 2 DOF control of the wrist; pressing the trackpad while the thumb is at the top, bottom, left, and right of the trackpad drove the hand to extend, flex, radially deviate, and ulnarly deviate, respectively. Pressing the trackpad in the intermediate space, say top left corner, would simultaneously rotate the wrist in both directions, in this case simultaneously extend and radially deviate. The trigger and grip button were used to supinate and pronate, respectively, and could be operated concurrently with the other 2 DOF.

Finally, we wanted participants to test the proposed novel trajectory control. In this mode participants only had access to two inputs, trigger and trackpad, similar to sequential control. In total there were five wrist trajectory modes, all of which rotate the hand along all 3 DOF simultaneously and had a predefined start and end point beyond which the wrist would not continue to rotate. Although all 3 DOF are used in each trajectory, these modes can be generally described as follows: *supination/ulnar deviation*, *flexion/ulnar deviation*, *supination/flexion*, *supination/extension*, *extension/ulnar deviation* (Fig. 5). Our version of trajectory control would cycle between these modes when both buttons (trigger and trackpad) were pressed, however, since only one mode is needed for each task, mode switching did not occur. At the beginning of each task a trajectory mode was preselected, and participants would only have to drive the wrist along a single trajectory to achieve the desired goal using either the trigger or the trackpad, depending at which end of the trajectory the task began. Each task had a single corresponding control mode (Table II) that would begin at either end of the trajectory, which was also predefined. Whichever orientation the wrist began at in this mode, would also be the start

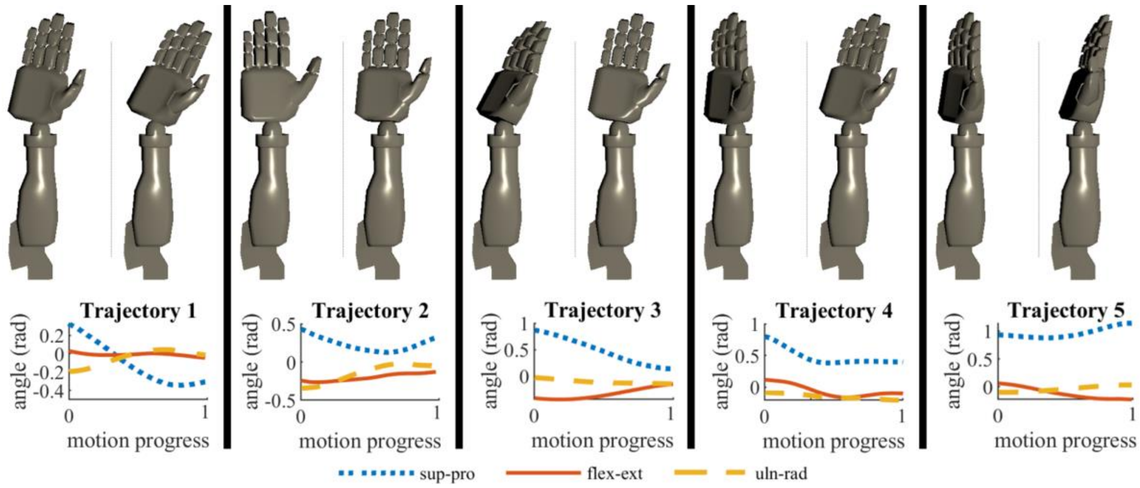


Figure 5. Each of the five wrist trajectory modes are depicted. First row represents the beginning and end (left to right) of each wrist trajectory respectively. Second row displays the actual 3 DOF angle trajectory. The order of rotations is supination-pronation, flexion-extension, and ulnar-radial deviation. Positive values correspond to supination, flexion, and ulnar deviation. Motion progress is scaled from 0 to 1, or 100% of the motion.

TABLE II. Task Description and Corresponding Trajectory Mode

Task description	Trajectory Mode Used
Reach to cup	4: supination-extension
Drink from cup	2: flexion-ulnar deviation
Reach to suitcase	5: extension-ulnar deviation
Transfer suitcase	4: supination-extension
Reach overhead	4: supination-extension
Bring can down	2: flexion-ulnar deviation
Reach to fork	5: extension-ulnar deviation
Use fork	4: supination-extension
Eat from fork	1: supination-ulnar deviation
Pour from cup	3: supination-flexion

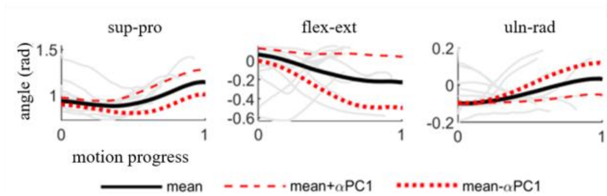


Figure 6. The recorded data that was used to create Trajectory 5 are shown in light grey. The trajectory was calculated by averaging the individual motions in the cluster. Variation in motion is represented by the first principal component (PC1), offset by either adding or subtracting it from the mean by one standard deviation (α).

orientation for the sequential and simultaneous control modes.

D. Trajectory modes

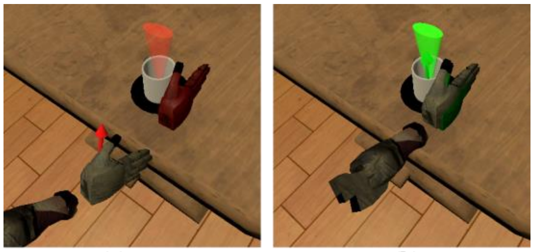
As mentioned earlier, trajectories were obtained in our previous work [5], and the method is summarized here. Using the same Vicon motion capture as in this study, we recorded the arm movements of 12 healthy subjects while they performed a set of ADL tasks, 24 in total. Wrist motions were segmented into 84 individual segments corresponding to reaching and transferring movement, which were clustered using data driven methods [25].

First, task repetitions were averaged using dynamic time warping (DTW) barycenter averaging (DBA). These averages, representing individual tasks, were then clustered using Hierarchical Clustering with Ward's distance measure, which groups data by maximizing the between group differences and minimizing the within group differences. The number of clusters was determined using the L method, which identifies the point of transition between the internally homogenous and non-homogenous clusters. Finally, each cluster was averaged using DBA to produce the five unique 3 DOF wrist trajectories (an example can be seen in Fig. 6). Determination of variation within each cluster was performed using functional principal component analysis (fPCA). We claim that these motions represent the full set of ADL tasks that prosthesis users would likely want their device to assist in performing.

E. Study Procedure

The role of an upper limb prosthesis is primarily to restore independent living by enabling amputees perform ADL tasks. The protocol therefore included a set of tasks inspired by work done in evaluating upper-limb prosthesis performance [30]. The virtual set up of the environment and objects are created to scale and are set up according to the dimensions described in [5]. We selected a set of 10 ADL tasks that require the use of only the right hand, cover a variety of locations, and include both reaching motion and object manipulation: (1) reach to cup, (2) drink from cup, (3) reach to briefcase, (4) transfer briefcase, (5) reach overhead, (6) bring down can, (7) reach to fork, (8) use fork, (9) eat from fork, and (10) pour. These were also selected in such a way as to span each of the trajectory modes, such that each mode was used at least once.

During reaching tasks participants were asked to begin with their hands relaxed by their side, and proceed with matching the orientation and location of the end effector, indicated with a semi-transparent red hand model (Fig. 7). Each task included a goal and tolerance for both the location and orientation; generally within 2 centimeters and 10°, inspired by previous pilot studies that we conducted. As such, task goals were defined agnostic to the control methodology and did not specifically align with any of the trajectory modes, and therefore it was possible that a given trajectory mode was insufficient and participants would need to use some amount of body compensation to ultimately reach the



(a) (b)

Figure 7. Example of a reach to a cup task. (a) Semi-transparent hand indicates the desired goal position of the user-controller hand, which dims as the hand approaches it. A red arrow is included next to the hand to assist with visualizing the current hand orientation. (b) Task completion occurs when the hand is within the Euclidean tolerance and the corresponding orientation arrow is within the tolerance cone.

desired goal. After a reaching task, the object was automatically placed within the hand, and participants were then asked to match the object location and orientation in a similar way, starting from where they grasped it. For sitting tasks, which included fork use and pouring tasks, participants were asked to place the hand in a relaxed position on the table prior to reaching. When using a control mode, the hand was reset to match the first frame of the trajectory control mode applicable to that task. Close attention was paid to how participants accomplished the task, such that the task was not deemed complete when participants unintentionally passed the hand through the goal, estimated based on how long the participant maintained the goal pose. Therefore participants were asked to repeat a task in which they could not maintain the goal pose for more than 1 second.

Tasks were performed in a semi-randomized order, with same-object reaching and manipulation occurring directly after one another. Each set of 10 tasks were performed in the following order: natural, no wrist, sequential, simultaneous, and trajectory control.

In order to mitigate training effects when faced with an unfamiliar environment or control interface, participants were given time to practice controlling the wrist prior to each task. Training ended when participants felt comfortable and had a strategy as to how they were going to accomplish the task, such that they were not hesitating when recording started. For trajectory control, a mode was preselected during training and placed at either end of the trajectory, according to the task requirements. During recording users would simply need to move along it to achieve the desired orientation.

F. Data Processing

Several metrics were considered in assessing whether there is a benefit to trajectory control. These include evaluating the range of motion (ROM) and Cartesian trajectory length of each joint as an indicator of natural movement, and were calculated for each control mode and task condition. Arm joints and torso coordinate systems were calculated according to [28]. Torso angles were calculated with respect to the pelvis and are described in the following order: torso flexion-extension (leaning forward or back), turning left-right (twisting), and leaning left-right, where extension, turning left, and leaning right are positive directions. Cartesian trajectory length, L, of each body segment was calculated as a sum of Euclidean distances between sampled points,

$$L = \sum_{i-1}^{j-1} \sqrt{(X_i - X_{i+1})^2 + (Y_i - Y_{i+1})^2 + (Z_i - Z_{i+1})^2} \quad (1)$$

where X, Y, and Z correspond to the three Cartesian components of a trajectory in space and j is the total number of sample points in that trial.

The start of the trial manually detected by the researchers when participants began to move, while the end was automatically determined by the software when the goal was reached. Time to accomplish the task was recorded for each task condition and would point to the simplicity of operating each control.

At the end of the experiment participants were asked to fill out a survey that included questions about their preferences and whether the control modes were easy to learn, intuitive, and appeared/felt natural.

III. Results

The time it took each participant to perform the set of tasks was recorded, and can be seen in Fig. 8. An aggregate time was assessed by summing across tasks for each control mode. The positive (natural) control was consistently the shortest for both participants, while sequential control took the longest. Of the three wrist controls, trajectory control was the fastest for both participants.

Range of motion of each joint angle was assessed to evaluate body compensation under different conditions and control modes (refer to Fig. 9 for the summary data, and Fig.

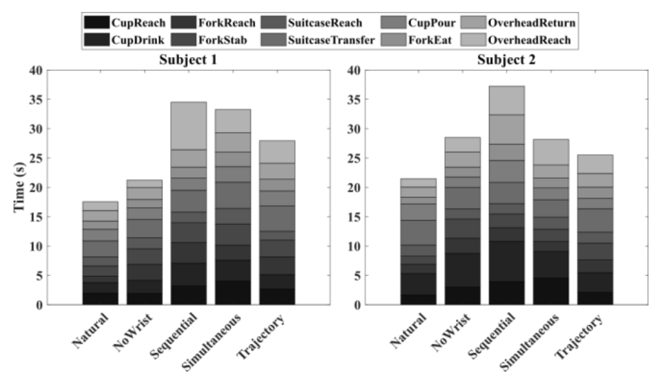


Figure 8. Time each subject took to perform the tasks using the five control modes. Each column represents a control mode and is broken up by individual task.

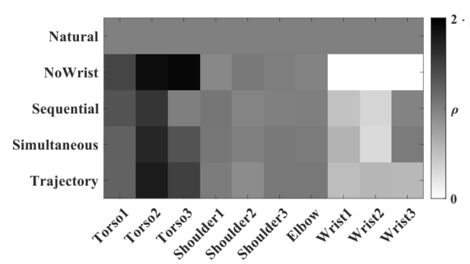


Figure 9. Average range of motion (ROM) results, for each condition across both participants, is displayed as a heat map, normalized for each column by the positive control (natural condition). Variable ρ represents the range of motion of the positive control. Joint angles are on the horizontal axis while control conditions are on the vertical axis.

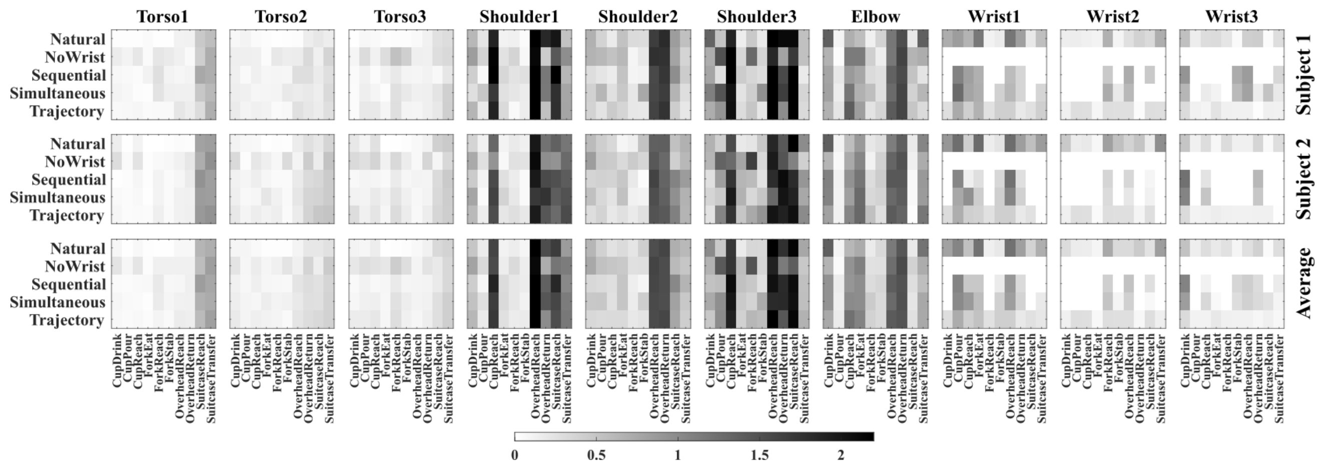


Figure 10. Range of motion (ROM) for each joint angle for each control mode and task is displayed, displayed as a heat map scaled to the largest angle (in radians). The first row corresponds to data obtained from the first subject, the second row corresponds to the second subjects, and the third row represents the average. Note that the wrist angles under the negative control (no wrist) condition is consistent, given that the wrist was fixed.

10 for the individual results). The negative control (no wrist condition) generally lead to larger torso and shoulder ranges of motion. While humeral elevation (Shoulder2 in the figures) was the lowest for trajectory control, other joints were higher. However, as a whole, results indicate very similar performance between trajectory control and the sequential and simultaneous controls. Additionally, negative control ("No Wrist" in the figures) had the highest range of motion for the torso angles and humeral elevation, consistent with expectation. Positive control (Natural in the figures) generally sustained the largest wrist angles range of motion, which likely contributed to the lowest torso and shoulder ranges.

Cartesian trajectory length of each joint is summarized in Fig. 11, averaged across both participants; displayed as a heat map. Each joint angle is normalized, from smallest to largest, to distinguish control mode performance. Total trajectory length is calculated by summing the lengths for each task while weighing them all equally; calculated as the sum of normalized lengths. Summing the unnormalized Cartesian lengths would result in a few tasks biasing the control mode performance. Positive and negative control dominated the two ends of the spectrum, while it was difficult to distinguish between the performances of the other three wrist modes.

After each experiment session subjects had the opportunity to give feedback by ranking various control qualities on a scale from 1 to 5 (Table III). Both subjects

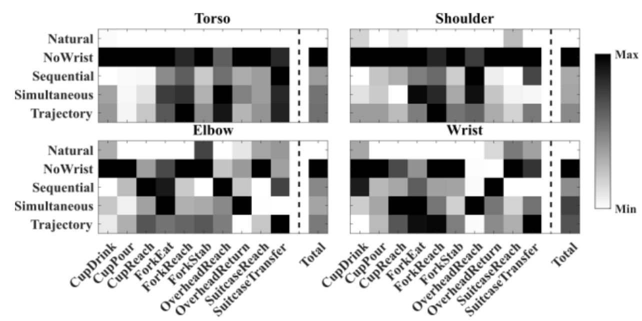


Figure 11. Cartesian trajectory length for each joint across each task and control condition. Vertical columns on the right summarize the results for each control condition. Columns are scaled independently from the smallest value (generally the Natural condition, positive control), to the largest value (generally the No Wrist condition).

ranked the positive control highest for training, intuition, cosmesis, and overall preference (full score of 5). Negative control received lowest marks for looking natural, and overall preference (1 or 2), however participants disagreed on whether training was easy or intuitive; average score of 3. Out of the three controllable wrist strategies, sequential performed the worst, while the trajectory control slightly performed simultaneous.

TABLE III. SURVEY RESULTS: AVERAGE SCORE

	Seq.[1]	Simul.[1]	Traj.[1]	No Wrist	Natural
Training was easy	4	4	4.5	3.5	5
Intuitive	1	4	4.5	3	5
Appeared natural	1.5	2.5	4	1.5	5
Overall preference	2	4*	4*	1	5

*One participant ranked one of the modes higher than the other; on average they received the same score.

[1]These are short for sequential, simultaneous, and trajectory control, respectively

IV. DISCUSSION

In this paper we assessed the performance of a novel wrist prosthesis control methodology, namely trajectory control, by comparing to alternative methods. Assessments included evaluating body compensation and cognitive load that users inevitably experience when faced with a complex orienting task. None of the participants reported an issue with achieving the end effector goal location and orientation, and felt that the tolerances were fair, even when struggling to complete certain tasks using the no-wrist control mode.

Participants were expected, but not required, to make use of the wrist control modes to assist them in completing the tasks. However, both participants consistently elected to use wrist functionality in every task with every DOF at some point being used throughout the experiment. This marks the benefits of a wrist prosthesis.

Participants did not use simultaneous control as expected, and largely operated the DOF sequentially. This is likely due to the difficulty associated with visualizing the interaction between different DOF and operating them simultaneously. However, because switching between modes was not required, task completion was nonetheless faster. This contributed to participants identifying this mode as easier to operate. This is consistent with previous findings [31]. Although possible, switching was not necessary for trajectory control, since a desired trajectory was preselected during training. This has the potential to bias results related to time. However, this was deliberately done so that we can demonstrate that orienting the hand using the novel control methodology is feasible, and perhaps superior to other control methods. In order to fully assess trajectory control, in future work we will further evaluate it by asking participants to complete ADL tasks that require switching and grasping.

Trajectory modes were faster compared to alternative control approaches, requiring users to operate a single DOF that automatically oriented the wrist close to the desired goal. Given that the five trajectories were create using a much larger set of motions, we did not expect the final hand orientation to be exact. However, body compensation, measured as ROM and Cartesian path length, was largely comparable to the other controls that had the capacity to orient the hand in any desired way. This suggests that the reduced cognitive burden is worth the reduction in direct orientation control. One interesting finding was that the no-wrist mode, was overall faster for one of the subjects. We suspect it was due to the reduced cognitive burden and perhaps due to the simplicity of the orienting strategy. Despite that, it had the largest amount of body compensation, appeared unnatural, and was the least preferred by both participants, suggesting that cosmesis is likely more important than speed.

While we did not quantify training time, we made a few observations. Training time for the natural and no wrist conditions was unsurprisingly very short, and time was primarily spent trying to simply understand what hand positions satisfied the task conditions. Out of the three wrist control modes, we observed that training time for the trajectory control was the fastest, as participants did not have to figure out a sequence of inputs. However, this is only the case if a mode is known in advance, as it was in our experiment. We suspect that if mode switching was required, as it will be in the real world where multiple trajectory sequences are used to complete a series of tasks, training time will increase and participants will have to focus on memorizing which modes are useful to which tasks.

Preliminary results demonstrate the ability of the trajectories to effectively carry out the tasks they're supposed to represent, by demonstrating that participants were able to complete all the tasks without major body compensation motions. This suggests that the modes could be practical in everyday use. To further validate their use, tasks would need to expand to motion beyond the ones that the trajectories constitute. It might be the case that the trajectories do not fully generalize and additional modes or fine tuning of the hand orientation will be required. Grasping, commonly the sole mode in powered transradial prosthetic devices, was also omitted as the focus was primarily on positioning the hand, but should be included in the future to assess whether the additional cognitive dissuade users from engaging with wrist control.

While we measured the cognitive load through various metrics, pupillometry has been shown to measure it directly [32]. It has been used within different fields, including prosthesis use [33] as well as driving [34]. Eye tracking will be implemented in the expanded version of this work by upgrading the HMD to include eye-tracking.

Dynamics too play a role in prosthesis use that is not currently captured in our set up. For full immersion, and to simulate real world prosthesis use, in future efforts we will expand our platform and task conditions to require object interaction, akin to [17]. In its present form controls are vastly simplified where the algorithms normally responsible for gravity and dynamics compensation are omitted. These will need to be developed when looking to transfer this technology to a real device.

Bimanual tasks were deliberately avoided in our experiment due to the lack of haptic feedback and the complexity associated with coordinating control with the healthy hand. It is the reason why amputees rarely perform bimanual interactions and why the research community is biased towards assessing unilateral tasks [35]. Given that one of the trajectories in [5] encompass a bimanual interaction, namely transferring a box, in the future we will test if trajectory control can assist with bimanual tasks as well.

Improvements to prosthetic wrist functionality need not to be limited to software control. Alternative mechanical wrist designs could simplify the types of controls that are needed altogether, for example, by focusing wrist rotation around a specific axis or trajectory. One such work includes rotating a single degree of freedom wrist device around an oblique axis. [36]; this axis was encompassed in one of the trajectories, namely the *flexion/deviation* trajectory mode.

Because rejection rates appear to be higher for more proximal levels of amputation [37], we believe that the application of this work will be even more valuable to elbow-wrist, and shoulder-elbow-wrist prosthetic devices [38]. We therefore plan to expand the VR platform to address these in future efforts using the additional trajectories identified in [5].

V. Conclusion

In the present work we highlight the benefits of a novel wrist control, based on 3 DOF trajectories, in completing daily tasks. We observed a vast reduction in cognitive burden associated with operating a 3 DOF wrist device while maintaining the benefits of a fully articulated wrist. Although participants did not have full control over the orientation of the hand, the much simplified control was nonetheless competitive with alternative approaches. These preliminary findings motivate the recruitment of more subjects and a larger investigation into wrist trajectory control as well as the development of a physical 3 DOF wrist device.

REFERENCES

[1] F. Montagnani, M. Controzzi, and C. Cipriani, "Is it Finger or Wrist Dexterity That is Missing in Current Hand Prostheses?," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 23, no. 4, pp. 600–609, 2015.

[2] A. J. Spiers, Y. Gloumakov, and A. M. Dollar, "Examining the Impact of Wrist Mobility on Reaching Motion Compensation Across a Discretely Sampled Workspace," *Proc. IEEE RAS EMBS Int. Conf. Biomed. Robot. Biomechatronics*, vol. 2018-Augus, pp. 819–826, 2018.

[3] C. R. Gambrell, "Overuse syndrome and the unilateral upper limb amputee: Consequences and prevention," *J. Prosthetics Orthot.*, vol. 20, no. 3, pp. 126–132, 2008.

[4] E. A. Biddiss and T. T. Chau, "Upper limb prosthesis use and abandonment : A survey of the last 25 years," *Prosthet. Orthot. Int.*, vol. 31, no. 3, pp. 236–257, 2007.

[5] Y. Gloumakov, A. J. Spiers, and A. M. Dollar, "Dimensionality Reduction and Motion Clustering during Activities of Daily Living : 3, 4, and 7 Degree-of-Freedom Arm Movements," *arXiv:2003.02641*, 2020.

[6] S. Ma and A. G. Feldman, "Two Functionally Different Synergies During Arm Reaching Movements Involving the Trunk," *J. Neurophysiol.*, vol. 73, no. 5, pp. 2120–2122, 1995.

[7] S. Wang *et al.*, "Evaluation of Performance-Based Outcome Measures for the Upper Limb: A Comprehensive Narrative Review," *PM R*, vol. 10, no. 9, pp. 951–962, 2018.

[8] N. M. Bajaj, A. J. Spiers, and A. M. Dollar, "State of the art in prosthetic wrists: Commercial and research devices," *IEEE Int. Conf. Rehabil. Robot.*, vol. 2015-Septe, pp. 331–338, 2015.

[9] W. Schweitzer, M. J. Thali, and D. Egger, "Case-study of a user-driven prosthetic arm design: Bionic hand versus customized body-powered technology in a highly demanding work environment," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, pp. 1–27, 2018.

[10] S. D. Iftime, L. L. Egsgaard, and M. B. Popovic, "Automatic Determination of Synergies by Radial Basis Function Artificial Neural Networks for the Control of a Neural Prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 13, no. 4, pp. 482–489, 2005.

[11] D. Blana, T. Kyriacou, J. M. Lambrecht, and E. K. Chadwick, "Feasibility of using combined EMG and kinematic signals for prosthesis control: A simulation study using a virtual reality environment," *J. Electromyogr. Kinesiol.*, vol. 29, pp. 21–27, 2016.

[12] R. R. Kaliki, R. Davoodi, and G. E. Loeb, "Prediction of Distal Arm Posture in 3-D Space From Shoulder Movements for Control of Upper Limb Prostheses," *Proc. IEEE*, vol. 96, no. 7, pp. 1217–1225, 2008.

[13] D. A. Bennett and M. Goldfarb, "IMU-Based Wrist Rotation Control of a Transradial Myoelectric Prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, no. 99, 2016.

[14] R. Volkmar, S. Dosen, J. Gonzalez-Vargas, M. Baum, and M. Markovic, "Improving bimanual interaction with a prosthesis using semi-autonomous control," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, pp. 1–13, 2019.

[15] A. Fougner, Ø. Stavdahl, P. J. Kyberd, Y. G. Losier, and P. A. Parker, "Control of Upper Limb Prostheses : Terminology and Proportional Myoelectric Control — A Review," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 5, pp. 663–677, 2012.

[16] S. Greengard, *Virtual Reality*, 1st ed. Cambridge, MA: The MIT Press, 2019.

[17] J. M. Lambrecht, C. L. Pulliam, and R. F. Kirsch, "Virtual reality environment for simulating tasks with a myoelectric prosthesis: An assessment and training tool," *J. Prosthetics Orthot.*, vol. 23, no. 2, pp. 89–94, 2011.

[18] B. N. Perry *et al.*, "Virtual Integration Environment as an Advanced Prosthetic Limb Training Platform," *Front. Neurol.*, vol. 9, 2018.

[19] L. Resnik, K. Etter, S. L. Klinger, and C. Kambe, "Using virtual reality environment to facilitate training with advanced upper-limb prosthesis," *J. Rehabil. Res. Dev.*, vol. 48, no. 6, pp. 707–718, 2011.

[20] M. Hauschild, R. Davoodi, and G. E. Loeb, "A Virtual Reality Environment for Designing and Fitting Neural Prosthetic Limbs," in *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 2007, vol. 15, no. 1, pp. 9–15.

[21] D. E. Levac, M. E. Huber, and D. Sternad, "Learning and transfer of complex motor skills in virtual reality: A perspective review," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, pp. 1–15, 2019.

[22] M. Kuttuva *et al.*, "The Rutgers Arm, a Rehabilitation System in Virtual Reality: A Pilot Study," *Cyberpsychology Behav.*, vol. 9, no. 2, pp. 148–151, 2006.

[23] B. N. Perry, C. Mercier, S. R. Pettifer, J. Cole, and J. W. Tsao, "Virtual reality therapies for phantom limb pain," *Eur. J. Pain*, vol. 18, no. 7, pp. 897–899, 2014.

[24] A. L. Alphonso *et al.*, "Use of a virtual integrated environment in prosthetic limb development and phantom limb pain," *Stud. Health Technol. Inform.*, vol. 181, pp. 305–309, 2012.

[25] Y. Gloumakov, A. J. Spiers, and A. M. Dollar, "A Clustering Approach to Categorizing 7 Degree-of-Freedom Arm Motions during Activities of Daily Living," in *IEEE International Conference on Robotics and Automation*, 2019.

[26] R. Davoodi, C. Urata, M. Hauschild, M. Khachani, and G. E. Loeb, "Model-Based Development of Neural Prostheses for Movement," *IEEE Trans. Biomed. Eng.*, vol. 54, no. 11, pp. 1909–1918, 2007.

[27] D. Putrino, Y. T. Wong, A. Weiss, and B. Pesaran, "A Training Platform for Many-Dimensional Prosthetic Devices Using a Virtual Reality Environment," *J. Neurosci. Methods*, vol. 244, pp. 68–77, 2015.

[28] G. Wu *et al.*, "ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion — Part II : shoulder, elbow, wrist and hand," *J. Biomech.*, vol. 38, no. 5, pp. 981–992, 2005.

[29] Y. Gloumakov, J. Bimbo, and A. M. Dollar, "Trajectory Control For a Myoelectric Prosthetic Wrist," in *Myoelectric Controls Symposium,* 2020.

[30] L. Resnik *et al.*, "Development and Evaluation of the Activities Measure for Upper Limb Amputees," *Arch. Phys. Med. Rehabil.*, vol. 94, no. 3, pp. 488–494, 2013.

[31] W. Zhang *et al.*, "Cognitive workload in conventional direct control vs. pattern recognition control of an upper-limb prosthesis," *Int. Conf. Syst. Man, Cybern.*, pp. 2335–2340, 2017.

[32] J. Beatty, "Task-Evoked Pupillary Responses, Processing Load, and the Structure of Processing Resources," *Psychol. Bull.*, vol. 91, no. 2, pp. 276–292, 1982.

[33] M. M. D. Sobuh *et al.*, "Visuomotor behaviours when using a myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 11, no. 1, pp. 1–11, 2014.

[34] O. Palinko, A. L. Kun, A. Shyrokov, and P. Heeman, "Estimating cognitive load using remote eye tracking in a driving simulator," *Eye Track. Res. Appl. Symp.*, pp. 141–144, 2010.

[35] A. Chadwell *et al.*, "Upper limb activity in myoelectric prosthesis users is biased towards the intact limb and appears unrelated to goal-directed task performance," *Sci. Rep.*, vol. 8, no. 1, pp. 1–12, 2018.

[36] A. Zinck, Ø. Stavdahl, E. Biden, and P. J. Kyberd, "Design of a compact, reconfigurable, prosthetic wrist," *Appl. Bionics Biomech.*, vol. 9, no. 1, pp. 117–124, 2012.

[37] L. V. Mcfarland, S. L. Hubbard Winkler, A. W. Heinemann, M. Jones, and A. Esquenazi, "Unilateral upper-limb loss: Satisfaction and prosthetic-device use in veterans and servicemembers from Vietnam and OIF/OEF conflicts," *J. Rehabil. Res. Dev.*, vol. 47, no. 4, pp. 299–316, 2010.

[38] L. Resnik, S. L. Klinger, and K. Etter, "The DEKA Arm: Its features, functionality, and evolution during the veterans affairs study to optimize the DEKA Arm," *Prosthet. Orthot. Int.*, vol. 38, no. 6, pp. 492–504, 2014.