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Non-Backdrivable Wedge Cam Mechanism for a Semi-Active Two-Axis Prosthetic Ankle

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Abstract: Frontal plane ankle motion is important for balance in walking but is seldom controlled in robotic prostheses. This article describes the design, control and performance of a semi-active two-degree-of-freedom robotic prosthetic ankle. The mechanism uses a non-backdrivable wedge cam system based on rotating inclined planes, allowing actuation only during swing phases for low power, light weight and compactness. We present details of the mechanism and its kinematic and mechatronic control, and a benchtop investigation of the system's speed and accuracy in ankle angle control. The two-axis ankle achieves angular reorientation movements spanning ± 10 deg in any direction in less than 0.9 s. It achieves a plantarflexion/dorsiflexion error of 0.35 ± 0.27 deg and an inversion/eversion error of 0.29 ± 0.25 deg. Backdriven motion during walking tests is negligible. Strengths of the design include self-locking behavior for low power and simple kinematic control. Two-axis ankle angle control could enable applications such as balance augmentation, turning assistance, and wearable perturbation training.

Keywords: prosthesis; ankle; non-backdrivable; inversion; plantarflexion; sagittal; frontal; two-axis; semi-active



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1. Introduction

Humans are less stable in the lateral direction than the anteroposterior direction during gait [1]. This instability is especially problematic for persons with amputation (PWA) of the lower limb, who lack control of the ankle in the frontal plane [2]. Existing commercially available prostheses address this problem only partially through passive mechanical compliance, such as split forefoot structural keels (e.g., [3,4]) or ankle bumpers, allowing for elastic inversion and eversion motion (e.g., [5]). This compliance remains inferior to the natural ankle's behavior, which can adapt its frontal angle to uneven ground and can even be used to actively correct balance perturbations through ankle inversion/eversion (IV/EV) control [6–9].

A few research prostheses have attempted to improve this shortcoming. Three include powered IV/EV control: one based on pure IV/EV angle control through an ankle module mounted above a standard prosthesis [10]; one that includes IV/EV under powered two-axis control using a cable system [11]; and one with fully powered two-axis control [12]. These mechanisms are effective in modulating IV/EV angle, but their height, mass and power consumption present challenges in deployment. An alternative approach is to accomplish a portion of the natural ankle's IV/EV function using a semi-active device—one that adjusts the passive properties of the prosthesis without supplying human-scale power to the movement. Semi-active approaches are popular in commercial prostheses with

sagittal plane adaptation, such as the robotic Össur *Proprio Foot*, [13], and hydraulic ankles such as Ottobock *Meridium* [14] and Triton *Smart Ankle* [15], Fillauer *Raize* [16], Proteor *Kinnex* [17], Endolite *Élan* [18], and College Park *Odyssey* [19,20]. Two such semi-active devices have been reported in research for frontal ankle motion: one that allows for free frontal motion during landing, then locks with a clutch to provide firm support [21], and one that controls frontal ankle stiffness through a variable-spring mechanism [22].

The aim of this paper is to describe the mechanics, design, control and performance of a non-backdrivable wedge cam mechanism for two-axis angle control, and its application to a novel semi-active ankle module called the Two-Axis 'Daptive Ankle (TADA) [23] that controls the ankle angle in both sagittal and frontal planes (Figure 1). Rather than adapting passively to different surfaces, the TADA mechatronically controls both plantarflexion/dorsiflexion (PF/DF) and IV/EV ankle angles. The inspiration for this concept is the natural ankle's ability to move in both directions, accomplished in the body by articulations at the talocrural and talocalcaneal joints. The TADA combines these movements into two-axis movement about a single joint center, exploiting the self-locking properties of the wedge cam mechanism to maximize torque holding capacity while minimizing the system's height and weight. Example intended use cases include matching ground slopes in arbitrary orientations; lifting or lowering the toes for stairs; inverting or everting the foot for turns [24,25]; and augmenting foot placement control with ground-matching IV/EV motion to enhance lateral balance [7,8].

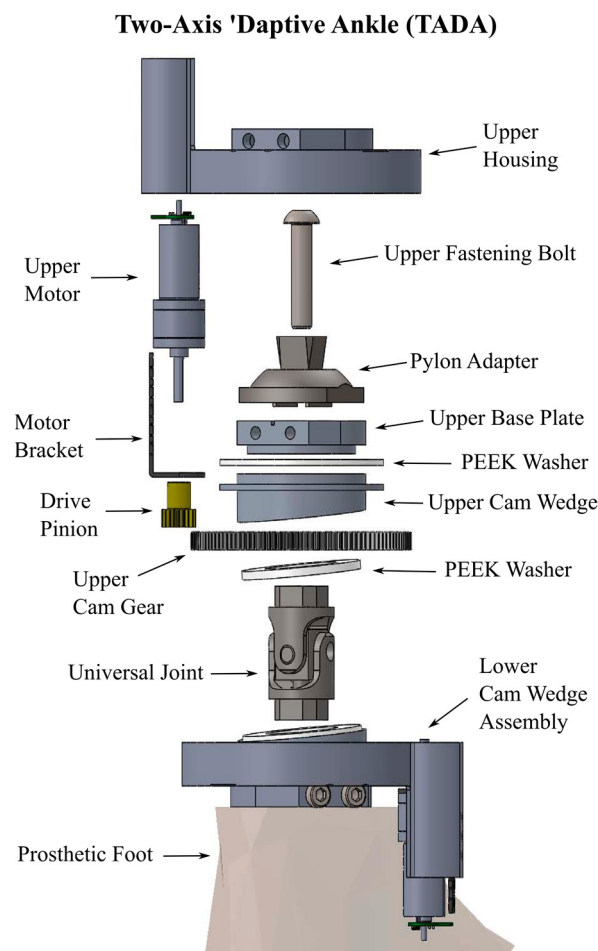


Figure 1. The TADA mechanism (upper half exploded) is made up of two angled wedge cams stacked on top of each other. By rotating the cams about the vertical axis, the ankle can achieve angles between zero and ten degrees in the frontal plane, sagittal plane or a combination of the two.

2. Materials and Methods

2.1. Mechanism Design Concept

The TADA aims to balance the convenience of passive ankles and the performance of powered ankles to add substantial function while minimizing added mass, height and power demand. Rather than actively powering ankle movements during stance phase [10–12], the TADA reorients the ankle during swing phases when the foot is off the ground. By adjusting the PF/DF and IV/EV angles, the TADA can alter the ankle moment experienced throughout the subsequent stance phase, thereby influencing whole-stride gait metrics such as the dynamic mean ankle moment arm [26]. The TADA is controlled based on readings from a foot-mounted inertial measurement unit (IMU) that allows reconstruction of foot movement to determine gait and some environmental conditions [27,28]. The TADA will then move to adapt to these conditions and/or to initiate or augment the body's response.

The TADA is based on a non-backdrivable wedge cam mechanism that can be driven by two small motors when unloaded but cannot move and is not backdriven when external loads (i.e., ankle moments) are applied. The mechanism consists of two stacked, cylindrical wedge-shaped cams that rotate about their cylindrical axes under the control of low-power DC motors. The mating faces of the wedge cams are cut at an angle of $\beta = 5$ deg with respect to the perpendicular cross-section of the cylinder. Rotation of the wedge cams changes the orientations of the two mating faces, and thereby also reorients the foot beneath. An internal universal joint (two-axis rotational joint) holds the stack together, supports axial and shear forces and prevents rotation of the distal prosthesis about the vertical axis. Each motor and pinion is mounted together with one of the wedge cams by a bracket and housing that ensure the motor and pinion stay in contact with the wedge cam's external gear all throughout the gait cycle. The mechanism is shown in Figure 1. The system can achieve angles of ± 10 deg in any combination of PF/DF and IV/EV. Critical design parameters are given in Table 1. Assembly images and videos of the movement are available in the Supplementary Materials.

Table 1. Dimensions for TADA components.

Parameter	Value
Universal joint—outer diameter	25.4 mm
Universal joint—radius to base of shear pin	8.40 mm
Wedge cam—outer diameter	50.0 mm
Wedge cam—inner diameter	42.0 mm
Wedge cam washer—outer diameter	46.0 mm
Wedge cam—face angle	5 degrees
Wedge cam—height at center of face	15.5 mm
Wedge cam—gear pitch diameter	70.0 mm
Wedge cam—number of teeth	120
Motor pinion gear—number of teeth	17

2.2. Strength Analysis of the TADA Mechanism

We designed the ankle for a hypothetical user with unilateral lower limb amputation with a body mass of $m = 100$ kg, foot length of 0.27 m (the most common prosthesis size), and a K2 Medicare Functional Classification Level (the most common level, indicating community ambulation but not highly dynamic activity). For this hypothetical case, we assumed the activity of interest was walking, and, therefore, we used design loads of 1.2 body weight (rounded to 1200 N) in the vertical direction and 0.2 body weight in the anterior and lateral directions (rounded to 200 N each) [29,30]. For structural analysis, we assumed a worst case in which the upper base plate is held fixed while all three of these loads are applied simultaneously at a ground contact point near the ball of the foot: roughly 55% of foot length (0.15 m) anterior from the ankle and 0.10 m below the ankle joint center (center of the U-joint). This assumption approximated loads in the late stance phase of

walking, shortly before toe push-off. We analyzed a neutral ankle angle configuration, neglecting the minor effects of changing cam wedge angles on contact loads. It is worth noting that these strength analyses were purely theoretical; the mechanism did not undergo rigorous destructibility nor durability testing beyond use in walking tests.

The critical components that determine the strength of the mechanism are the pins in the central universal joint and the surfaces of the wedge cams. The wedge cam is subjected to compressive stress on one side at a time to support the ankle joint moment arising from the ground contact force. To analyze the stress on the wedge cam, we computed a moment balance of the sagittal applied force about the ankle, neglecting the small inversion/eversion moment produced by the lateral force. Thus, a 1200 N vertical load together with a 200 N anterior load creates a 200 Nm plantarflexion torque T_{ank} supported by the ankle. To analyze the load on the U-joint, we assumed well-aligned contact between the wedge cams, implying by Hooke's law that the cam surface stress varies linearly in proportion to the distance from the ankle joint, up to the wedge cam radius r . In that case, the resultant reaction force through the wedge cam face has a moment arm of $\pi r/4$ about the ankle joint (0.0172 m), leading to a contact force magnitude of 11,600 N.

The peak contact stress depends on the radial width t of the wedge cam washer that is supported by the cam wall: $\sigma_{\text{max}} = 2T_{\text{ank}} / (\pi r^2 t \cos \beta)$. For the TADA as built, $t = 0.002$ m and $r = 0.022$ m (midpoint radius of the portion of the washer supported by the wedge cam wall), so the maximum stress is estimated to be $\sigma_{\text{max}} \approx 132$ MPa. For the PEEK material used at this interface, the permissible static surface pressure is 150 MPa or higher, so the estimated stress is within this strength limit. A more conservative (worse-case) estimate is to assume small-region contact at the nominal radius of the wedge cam. In this case, the moment balance about the ankle yields a smaller contact force of 9100 N due to the greater moment arm (0.022 m). But the smaller contact region (estimated at 0.015 m tangential by 0.002 m radial) yields an estimated stress of 300 MPa, which would be well above the material limit. Therefore, strength requirements suggest that the system needs to be well-aligned to distribute the contact stress as in the nominal case. These same stress conditions apply to the other PEEK washers between the wedge cams and the base plates. In use, none of the PEEK washers experienced damage, suggesting that the conservative approach may overestimate the stress and the true stress is likely between the two estimates given.

The U-joint supports forces and moments in all directions except the two rotational degrees of freedom: it supports axial tension to balance the compressive stress on the wedge cam; shear forces to prevent translation of the foot; and axial torque to prevent the foot from undergoing internal/external rotation. To analyze the load on the U-joint, we used the applied forces and the cam reaction force estimated above using the linear stress distribution on the cam surface. Both the tensile and the torsional loads on the U-joint are physically supported by nearly pure shear forces in the pins of the U-joint, in two orthogonal directions (anteroposterior, mediolateral). The estimated tensile load on the U-joint due to the 1200 N vertical forefoot load and the ankle joint moment is 10,400 N, or 5200 N vertical shear force per pin. The estimated horizontal components of shear in the pins include contributions from both direct-acting mediolateral and anteroposterior forces and the torsion produced by the mediolateral force on the ball of the foot. With a radius to the base of each pin of 0.0084 m, static equilibrium analysis yields horizontal shear forces of 3500 N in two pins and 3700 N in the other two pins. The resultant of vertical and horizontal shear force (maximum 6350 N) can be used directly with a maximum shear stress criterion to design the pin components in a custom U-joint; it is equivalent to a pure torque of 54 Nm on the U-joint. The U-joint used in the TADA was instead chosen as an off-the-shelf component (1-inch (0.0254 m) diameter steel U-joint, McMaster-Carr, Atlanta, GA, USA) to exceed this strength, as determined by its maximum torque rating (344 Nm).

Final design parameters for the TADA as built are shown in Table 1. According to the analysis above, the ideal (close contact) case has a safety factor of roughly 1.14 on the PEEK material and 6.4 on the U-joint. These safety factors allow for imperfect device alignment and/or more dynamic activities.

2.3. Ankle Kinematic Control

Kinematic and inverse kinematic laws for controlling ankle angle through wedge cam angular position are derived using a serial-chain manipulator approach. Reference frames (Figure 2) are placed on the upper base plate at its bottom face (frame 0; axes x_0, y_0, z_0 , attached rigidly to the prosthetic pylon/shank); the upper wedge cam on its top orthogonal face (frame 1) and bottom inclined face (frame 2); the lower wedge cam on its top inclined face (frame 3) and bottom orthogonal face (frame 4); and the lower base plate on its top face (frame 5; attached rigidly to the foot). The upper and lower wedge cams are driven to rotate about the z_0/z_1 axis and the z_4/z_5 axis, respectively, and the wedge faces are cut at a fixed angle β (here 5 degrees) about the y_1/y_2 and y_3/y_4 axes.

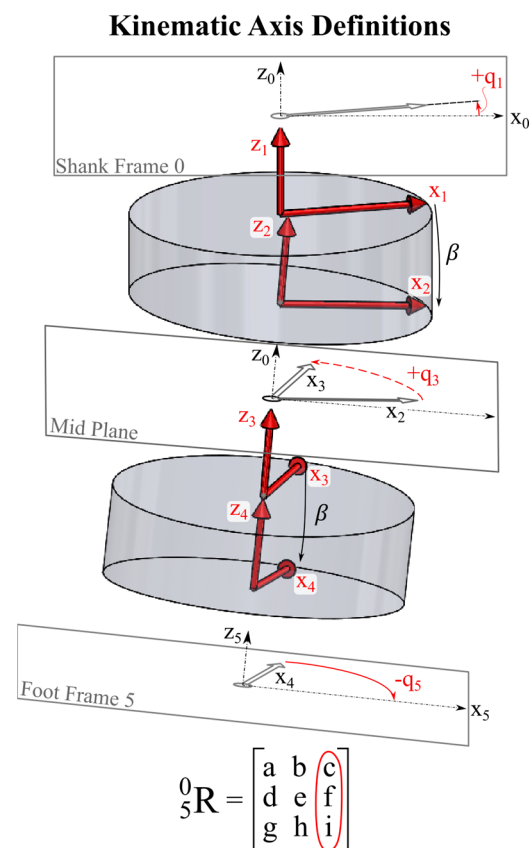


Figure 2. CAD model of TADA, displaying reference frames 0 to 5 used to derive kinematic control laws. Each rotation is positive when the distal component rotates about its respective $+z$ axis relative to the proximal component, according to the right-hand-rule. All components distal to a given rotation are assumed to move with the rotating part for analysis; the final rotation q_5 is used to enforce the constraint that the toes point forward (1). The matrix 0_5R built from these rotations is used to deduce the control parameters, notably the last column $[c \ f \ i]^T$, which represents the axis z_5 in frame 0 (see Figure 3).

The individual rotations are subject to the constraint that the toes point forward, and they have a complex relationship to final PF/DF and IV/EV angles. They can be described by a sequence of rotations relative to the tibial pylon axes (frame 0): first a controlled rotation q_1 of the first wedge about z_1 ; then a fixed rotation of angle β about axis y_2 ; then a rotation q_3 of the second wedge about axis z_3 ; then a fixed rotation β about axis y_4 ; then a final controlled rotation of the foot itself by an angle q_5 about axis z_5 . This final rotation keeps the foot facing forward, a constraint which is enforced physically by the U-joint and described mathematically below (see Equations (4)–(8)). The composition of all these rotations yields a complicated rotation matrix 0_5R expressing the orientation of

frame 5 in coordinates of frame 0 (see Appendix A). This rotation matrix can be used to solve for the rotations necessary to achieve a desired foot orientation. In this approach, the desired PF/DF and IV/EV angles are used to compute a desired rotation matrix (R_{PFIV} ; see Appendix A), and terms are matched to the equivalent rotation matrix 0_5R to determine the required wedge cam rotation angles q_i . These equations are complex and require numerical solution.

However, given the constraints of the U-joint and wedge cams, we chose to parameterize the ankle angle control to be based on two inputs of (a) the magnitude of the tilt angle, θ , and (b) the “downward direction” relative to the pylon, α (see Figure 3). θ represents the tilt magnitude (0–10 deg) of the foot’s z_5 axis from the pylon’s z_0 axis. α represents the direction of the projection of the foot’s z_5 axis on the x_0 - y_0 plane, measured as an angle from the x_0 axis (−180 to 180 deg). This formulation represents the 3D orientation of the foot more intuitively, especially for tasks such as matching a world-frame ground incline (in which the incline’s magnitude and direction are known, but PF/IV angles are not obvious).

Downward Direction Control

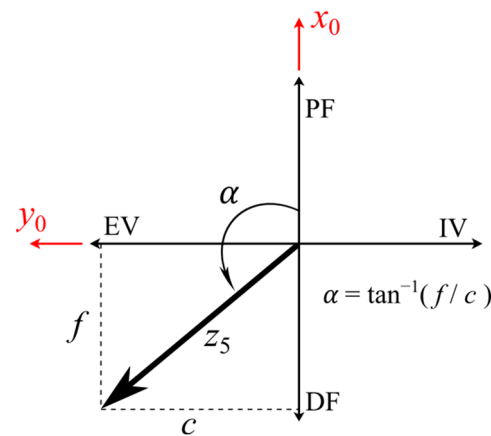


Figure 3. Projection of desired z_5 axis of the foot frame into the leg fixed frame (frame number 0) determines a “downward” direction defined by angle α . For example, setting α to 135 degrees yields an equal combination of DF and EV. f and c refer to elements of the desired rotation matrix 0_5R , see Figure 2.

The (θ, α) pair can be used to create an axis–angle formulation of the rotation matrix (R_{ax_ang} ; see Appendix A) that can be matched numerically to 0_5R as above. Alternatively, a closed-form relationship can be derived. In the rotation matrix 0_5R , the third column represents the unit vector z_5 (the surface normal to the foot’s top face) expressed in the pylon reference frame (frame 0): ${}^0z_5 = [c, f, i]^T$ (see Figure 2). The last element i is the projection of z_5 onto the z_0 axis: simply the cosine of the overall foot inclination angle θ . To control foot inclination angles (ranging 0 to 2β), the last element i is compared to the symbolic rotation matrix 0_5R (Appendix A):

$$i = \cos(\theta) = \cos^2(\beta) - \cos(q_3) \sin^2(\beta). \quad (1)$$

Thus, the relative rotation angle between the upper and lower wedge cams, q_3 , can be computed in closed form and simplified:

$$q_3 = \cos^{-1} \left(\frac{\cos^2(\beta) - \cos(\theta)}{\sin^2(\beta)} \right) = 2 \cos^{-1} \left(\frac{\sin(\frac{\theta}{2})}{\sin(\beta)} \right). \quad (2)$$

The first two elements of 0z_5 (c, f) are the projection of z_5 (the foot frame “up” axis) into the (x_0, y_0) plane; this projection can be viewed as a direction vector indicating which part

of the foot is intended to point most downward relative to the pylon. This vector direction is defined by the vector's angle α from the $+x_0$ axis (Figure 3). This direction can be deduced from the target rotation matrix if PF/DF and IV/EV angles are specified, or directly as a downward direction vector if this is known, e.g., from the slope of a known ground surface relative to the leg. The angle α is computed from the four-quadrant arctangent:

$$\alpha = \tan^{-1}(f/c). \quad (3)$$

For the maximum inclination angle ($\theta = 2\beta$) in any direction, α is the angle at which the thickest parts of the wedge cams are aligned on top of each other, pushing that side of the prosthesis downward and lifting the opposite side. In this case, $q_1 = -q_5 = \alpha$ and $q_3 = 0$. In the current control algorithm, $\alpha = \{0, 180\}$ degrees correspond to pure PF/DF, while $\alpha = \{-90, 90\}$ degrees correspond to pure IV/EV (Figure 3).

For the “neutral” configuration ($\theta = 0$), q_3 is 180 deg and q_1 and q_5 are each 90 deg from α (Figure 4). In this case, the equal and opposite face inclines cancel to hold the mechanism in the shape of a cylinder. It can be observed that the neutral configuration is not unique; it exists with equivalent effect for all α .

Neutral Pose

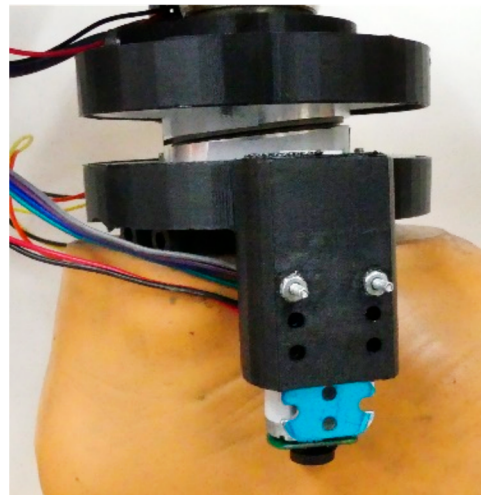


Figure 4. The TADA ankle device in neutral position. Neutral PF/DF and IV/EV angles are achieved by aligning the top and bottom wedge cams in opposite orientations so their respective face angles cancel. This pose is not unique; the two cams may be rotated together without changing foot orientation.

For intermediate angles ($0 < \theta \leq 2\beta$), the upper and lower wedge cams are controlled to rotation angles q_1 and q_5 by separate motors. These angles are determined by the downward direction α and the rotation q_3 between the wedge cams, which are each an equal rotation from α in opposite directions. This relationship is captured by two equations:

$$q_1 = \alpha - \tan^{-1}\left(\frac{\tan\left(\frac{q_3}{2}\right)}{\cos(\beta)}\right), \quad (4)$$

$$q_5 = -\left(\alpha + \tan^{-1}\left(\frac{\tan\left(\frac{q_3}{2}\right)}{\cos(\beta)}\right)\right). \quad (5)$$

The second terms in (4) and (5) account for the face angle β , which distorts the effects of target rotations at the inclined face (q_3) on the actuated rotations at the orthogonal

faces (q_1 and q_5). Because β is a small angle, simplified equations are adequate for more intuitive control:

$$q_1 \approx \alpha - q_3/2. \quad (6)$$

$$q_5 \approx -(\alpha + q_3/2). \quad (7)$$

These simplified equations are accurate within 0.11 degrees of cam rotation for the current design ($\beta = 5$ deg), leading to negligible ankle angle error. They also imply an intuitive relationship among the three rotations that describes conceptually how q_5 acts to turn the toes forward after rotations q_1 and q_3 :

$$q_1 + q_3 + q_5 \approx 0. \quad (8)$$

Note that q_3 in (2) results from the inverse cosine function, and hence has both positive and negative solutions; the sign determines which wedge cam turns clockwise and which counterclockwise from the downward direction angle α , but the effects are equivalent. Also note that q_5 is specified as a positive rotation of the foot relative to the lower wedge cam about axis z_5 (see Figure 2), but is actually implemented by driving the lower wedge cam to an angle $-q_5$ relative to the lower base plate/foot, where the motor is grounded. This relationship addresses the seemingly counterintuitive sign of q_5 in (7). Finally, it is important to note that despite the U-joint's action of resisting axial rotation, the multiaxial movement still allows the x -axis of the foot frame (x_5) to deviate slightly from the sagittal plane. With a face angle of 5 deg, the maximum deviation is 0.43 deg, occurring when the downward direction angle α is in the set $\{\pm 45, \pm 135\}$ deg.

2.4. Nonbackdrivability of the Wedge Cam Mechanism

The wedge cam mechanism uses friction to prevent the cams from backdriving during stance phase. This is a critical component of the semi-active design: it allows for the use of small motors and transmission components because they do not support large external moments from body weight. The inclined surface between the two wedge cams experiences a compressive load distributed over one side of the contact area, which forms a couple with the tensile load in the universal joint to support the ankle moment (Figure 5A). The compressive load on the wedges, acting at an inclined angle (Figure 5B), creates a twisting moment on each wedge, which is held in balance by friction on both the angled face and the orthogonal face (Figure 5C). The worst case is neutral PF/DF (downward direction angle $\alpha \in \{0, 180\}$ degrees with the two wedges' x_1 and x_3 axes toward the left and right) with a large vertical force applied at the forefoot to create a large ankle moment M_{ankle} . In this case, the contact stress on the inclined face between the two wedge cams creates a backdriving twist moment M_{back} on each:

$$M_{\text{back}} = M_{\text{ankle}} \tan(\beta). \quad (9)$$

Fortunately, this twist moment is countered by the moment capacity of friction on both the upper and lower surfaces. Analysis of these balancing moments under a Coulomb friction model reveals a relationship between the surface friction coefficient and the incline angle of the wedge cams:

$$\tan(\beta) \leq \frac{2\mu}{1 - \mu^2}. \quad (10)$$

This relationship indicates that a friction coefficient as low as 0.05 is sufficient to ensure locking at a face inclination angle of up to 5.7° . The true friction coefficient is expected to be higher ($\mu \approx 0.1$ for the PEEK plastic used at the interface, allowing angles up to 11.4°); therefore, the current face angle of 5.0° is well within the safe, non-backdrivable range. Friction also opposes rotation of the wedges by the motors during swing phases, but since the normal force is very low in that case, it is easily overcome by small motors. Detailed analysis of the friction locking mechanism is presented in Appendix B.

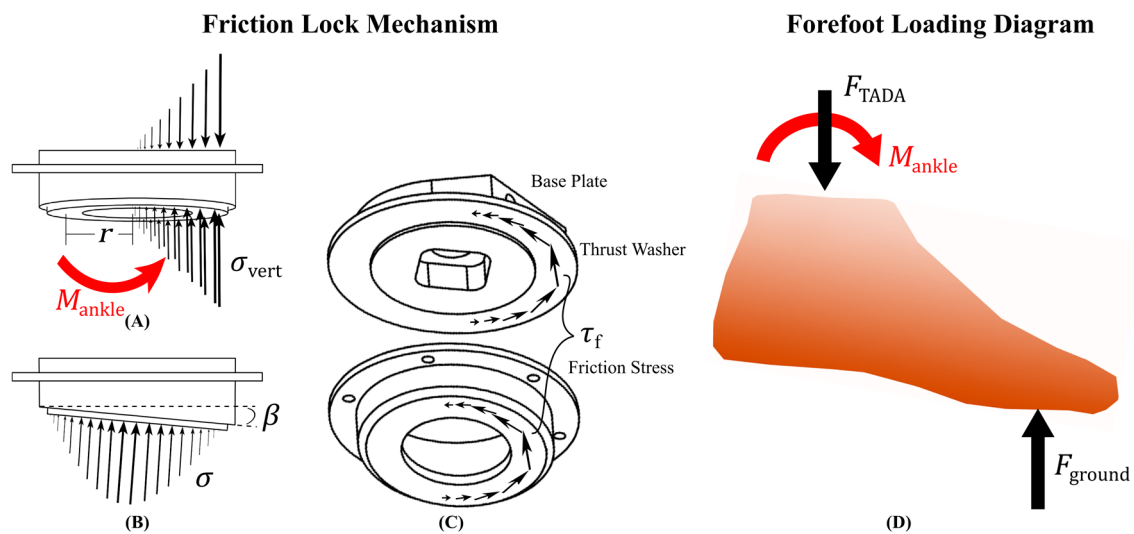


Figure 5. Stresses on the upper wedge cam in the worst-case loading scenario for backdriving of the rotational mechanism: vertical loading on the forefoot (denoted here by M_{ankle}), with the wedge cams' x-axes oriented in the frontal plane. (A) Side view of the upper wedge cam showing the distribution of vertical-direction stresses σ_{vert} . Only the front half of the wedge cam is loaded, compressed from below by the lower wedge cam and from above by the upper base plate. The toes are pointing to the right in this view. (B) Front view showing sinusoidal distribution of the normal reaction stress σ on the inclined face. The toes are pointed out of the page in this view. (C) Friction stress capacity τ_f is determined by the coefficient of friction and the normal stresses. Friction from both surfaces prevents backdriving of the mechanism by the inclined normal stress. (D) Sagittal view of the free body diagram of the prosthetic foot. The force of the ground on the foot (F_{GROUND}) acts on the forefoot to create counteracting reactions F_{TADA} and M_{ankle} at the ankle joint.

2.5. Control of the Ankle Mechanism

2.5.1. Sensing and Actuation for TADA Control

Small brushed DC gear motors (31:1 Metal Gearmotor 20Dx41L mm 12V CB; Pololu Corp., Las Vegas, NV, USA; mass 44 g each) are mounted in the upper and lower housings to actuate the two wedge cams. Each motor has an acetal pinion gear (17 teeth) mounted on its output shaft, which drives a spur gear (120 teeth) mounted on the wedge cam to rotate it during swing phases (SDP/SI, Hicksville, NY, USA). Each motor includes a rotational quadrature encoder for position control (375 counts per revolution of the pinion). The gearmotors have nominal stall torque of 0.24 Nm at 1.6 A drive current, and no-load speed of 47 rad/s (450 RPM) at 12 V applied voltage; assuming no friction, this no-load speed would cause the wedge cam to rotate at 6.66 rad/s. With the maximum required angular reorientation being $\pm\pi$ radians (± 180 deg), this movement speed yields a movement time estimate of 0.47 s. In reality, the friction at the interfaces restricts this motion, and we expect the motor to run near its peak power speed (roughly half the no-load speed). This condition predicts wedge cam movement times of roughly 0.9 s for ± 180 deg reorientation.

The motors are controlled by a Raspberry Pi 3B embedded computer, together with a dual H-bridge motor control board (MAX14870; Pololu Corp., Las Vegas, NV, USA), a dedicated 8-bit microcontroller for encoder counting and analog sensor reading (Alamode; Wyolum, Reston, VA, USA), and two analog bipolar Hall effect sensors to read permanent magnets installed on the rotating wedge cams to set the "home" position (DRV5053, Texas Instruments, Dallas, TX, USA). An inertial measurement unit (IMU; 3 Space; Yost Engineering, Portsmouth, OH, USA) is mounted on the foot to distinguish stance and swing phases in the gait cycle and to reconstruct foot trajectory [27] to support control decisions [28]. The embedded computer is programmed in Python (for logic and motor control) and the microcontroller in C (for sensor management).

The rotation of the wedge cams is controlled using a shifted bang–bang control law for the motors (Figure 6). Bang–bang control means that the motor receives either full voltage (maximum allowed PWM duty cycle, forward or backward) when its position is outside a deadband surrounding the target position, or no voltage at all (in this case, short-circuit braking mode) [31]. This style of control is chosen because the friction-based mechanism prevents smaller inputs (lower PWM duty cycle) from moving the wedge cams at all. More traditional algorithms like proportional control would continually apply an ineffectual small current to the motors when near but not perfectly at the target position, wasting battery power and heating the motors. The shifted bang–bang controller adds a direction-based component to the normal bang–bang scheme: the deadband is shifted toward the side from which the angle is approaching, so that the command is set to zero earlier (Figure 6) and rotation stops closer to the target. The parameters are the width and shift of the deadband and the amplitude of the active signal. Shifted bang–bang control reduced overshoot, improved positioning accuracy and eliminated residual power consumption relative to other control schemes tested (results are shown in Figures 7 and 8).

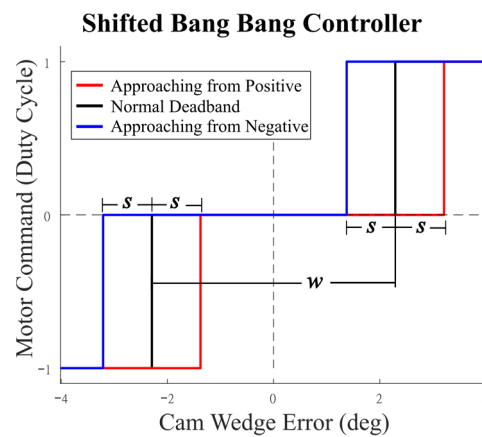


Figure 6. Shifted bang–bang control law (blue and red), compared to standard bang–bang control (black). Shifting the deadband based on the approach direction reduced overshoot. The deadband is parametrized by width w and shift s in units of angle.

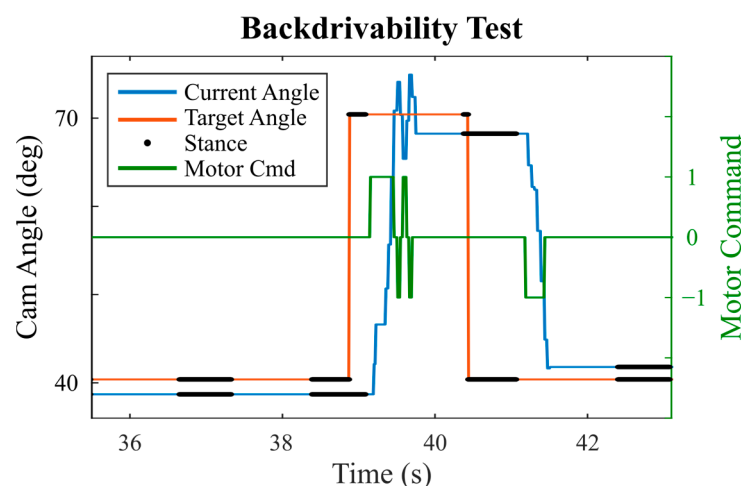


Figure 7. Example of cam movement during swing phases and non-backdrivability during stance phases. The target angle (red) can change anytime, but the cam angle (blue) moves only when actuated by the motor (green, right axis) during swing phases. The cam angle does not change when the ankle bears body weight during stance phases (black). For motor commands, ± 1 indicates maximal commands, and 0 is off.

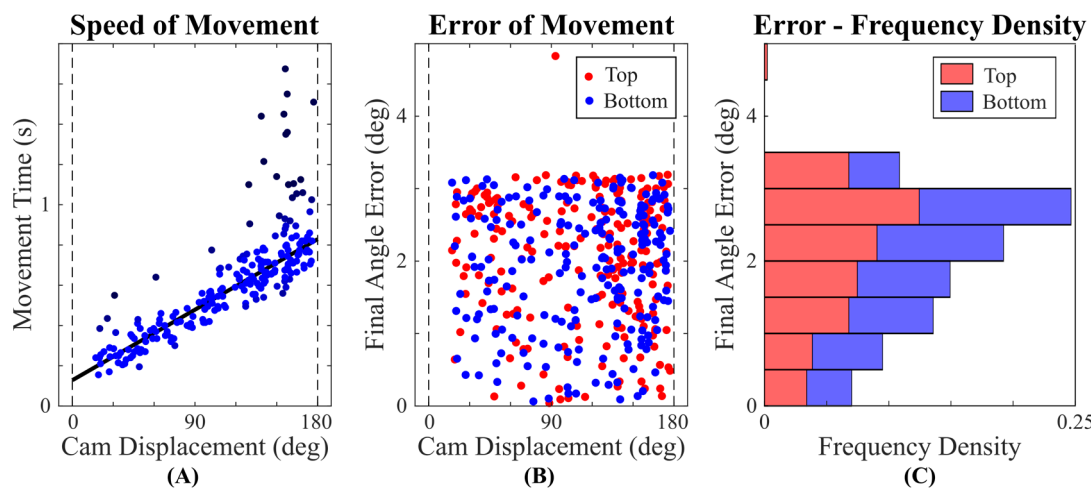


Figure 8. (A) A weighted least-squares fit suggests a linear relationship between wedge cam displacement and movement time ($t = 0.0039 \Delta\theta_{\text{top}} + 0.1282$). Darker color indicates outliers with low weighting. (B) Error of movement displays that all but one of the wedge cam displacements have final errors below 3.2 degrees. Outliers in both plots indicate instances when the mechanism was briefly stuck. (C) Frequency density of observed final wedge cam angle errors.

2.5.2. Performance Optimization

There are two numerical indeterminacies in solving for the required wedge cam orientation angles, which provide some freedom to optimize performance based on the specific circumstances of every angle adjustment. First, q_3 could be either positive or negative due to the inverse cosine function used to define it. The difference determines which wedge cam is more clockwise than the other, but the two solutions produce equivalent ankle angles. Therefore, we include logic to choose the sign of q_3 that minimizes the movement time from the prior orientation setting, i.e., the choice that results in less excursion of the two wedge cams. Second, the “downward” direction α is indeterminate whenever the desired inclination angle is 0. This neutral ankle pose is achieved by any combination of wedge cam orientations that are 180 degrees apart (Figure 4). Therefore, when returning to neutral, we include logic that moves directly toward the “nearest neutral” from the prior setting, by maintaining the last α value and setting q_3 to 180 degrees.

2.5.3. Safety Precautions to Eliminate Unnecessary Motions of the TADA

Certain safety precautions are also necessary during the movement. For large changes in ankle pose, the intermediate ankle angles that occur during the movement of the wedge cams can sometimes be very different from either the new or the old setting; the ankle may look like it is drawing a circle with the toes as it moves. When these intermediate poses include plantarflexion, the toes could contact the ground and trip the user. Therefore, we implement an interpolation method that eliminates this undesired “toe circumduction” phenomenon by interpolating the total motion into many small steps and issuing commands for all the intermediate steps at a rate both motors can achieve. First we determine the angle through which each wedge cam must rotate, and estimate the total movement time based on the typical movement speed. Then we interpolate both cams’ angle changes into n pieces at 10 Hz spanning the estimated movement time. Finally, we issue the interpolated commands to both motors at 10 Hz to maintain synchrony. This interpolation method successfully eliminates the hazardous toe movements.

2.5.4. Performance Testing of the Accuracy and Backdrivability of the TADA

To test the backdrivability of the TADA under external loads, we performed walking trials on the TADA with a standard low-profile prosthesis (Seattle Natural Foot; Trulife, Dublin, Ireland) using a prosthesis simulator boot on the right leg of an unimpaired user (a

member of the research team) and a contralateral lift shoe [32]. An experimenter manually commanded the TADA to alternate between a specific inclined posture and a neutral posture several times within each one-minute walking trial. We performed a trial for each of the nine angle settings: 5 and 10 deg settings in each of PF, DF, IV and EV, and a neutral-only trial. We recorded data from the onboard prosthesis controller at 84 Hz (observed controller frequency). These controller data included the angular position of the top and bottom wedge cams, their respective target angles, and the top and bottom motor commands. We also recorded the angular velocity of the foot from the attached IMU. We quantified backdrivability as the total range of motion of the top and bottom wedge cams during stance periods within each trial (results in Figure 7).

To validate the design and control approach, we tested ankle angle control performance with an external motion capture system for reference. We attached the TADA to a mounting frame and attached a standard foot prosthesis to the TADA. We commanded the prosthesis to move to 48 poses five times each in a random order (240 total random movements), with a dwell time of 2.5 s at each orientation (see Figure 9, the command poses are the white circles with red outlines). We measured foot motion using optical motion capture (12-camera Optitrack Prime 13 system; NaturalPoint, Corvallis, OR, USA), with five reflective markers attached to the foot prosthesis and five attached to the mounting frame.

We calculated the performance of the wedge cams in reorienting quickly and accurately (results in Figure 8). We calculated the angular displacement of each wedge cam movement from its starting pose to each new target, and used motion capture data to measure the time from the first ankle movement (2 deg/s threshold) to the first arrival within 0.1 deg of the final pose (excluding fine settling adjustments). We plotted movement time vs. displacement and performed a recursive weighted least squares fit to determine a linear relationship between them (MATLAB R2021a *fitlm* function with robust fitting), to inspect the typical and worst-case movement times. We also recorded the error of the final wedge cam orientations relative to their commanded orientations. We plotted error vs. displacement to inspect the accuracy achieved by the shifted bang-bang controller. We further plotted a frequency density histogram of movement errors to observe the distribution of final wedge cam orientation errors. We calculated whether the distributions were Gaussian using Kolmogorov–Smirnov and Anderson–Darling normality tests (MATLAB R2021a functions *kstest* and *adtest*).

We analyzed the motion of the prosthesis using inverse kinematics software (Visual3D v6.01.25; C-Motion, Germantown, MD, USA) and custom scripts in MATLAB R2021a. To match observed movements to commanded movements despite potential misalignment during assembly, we created a functional ankle reference frame from the functional sagittal and frontal axes of the prosthetic ankle movement. First we defined a functional joint center (FJC) using a subset of the 240 random movements, analyzed with the Gillette method built into Visual3D [33,34]. Next, we found the functional sagittal axis using only PF/DF movements, and the functional frontal axis using only IV/EV movements. The two functional axes are orthogonal and intersect at the FJC.

We used the inverse kinematic model to compute the PF/DF and IV/EV angles about these functional joint axes from the stationary posture following each of the 240 movements. We compared the commanded and measured angles of the prosthesis. We computed the mean and standard deviation of the five repetitions at each pose to characterize repeatability (results in Figure 9A).

Finally, we characterized backlash in the kinematic mechanism for a subset of commanded poses. We commanded several poses (−10, −6.67, −3.33, 0, 3.33, 6.67, and 10 degrees of pure PF and pure IV) and manually rotated the prosthesis in a clockwise motion within the backlash limits in both directions. We recorded the maximum deviation in each direction as an indication of how tightly the prosthesis maintains each posture (results in Figure 9B).

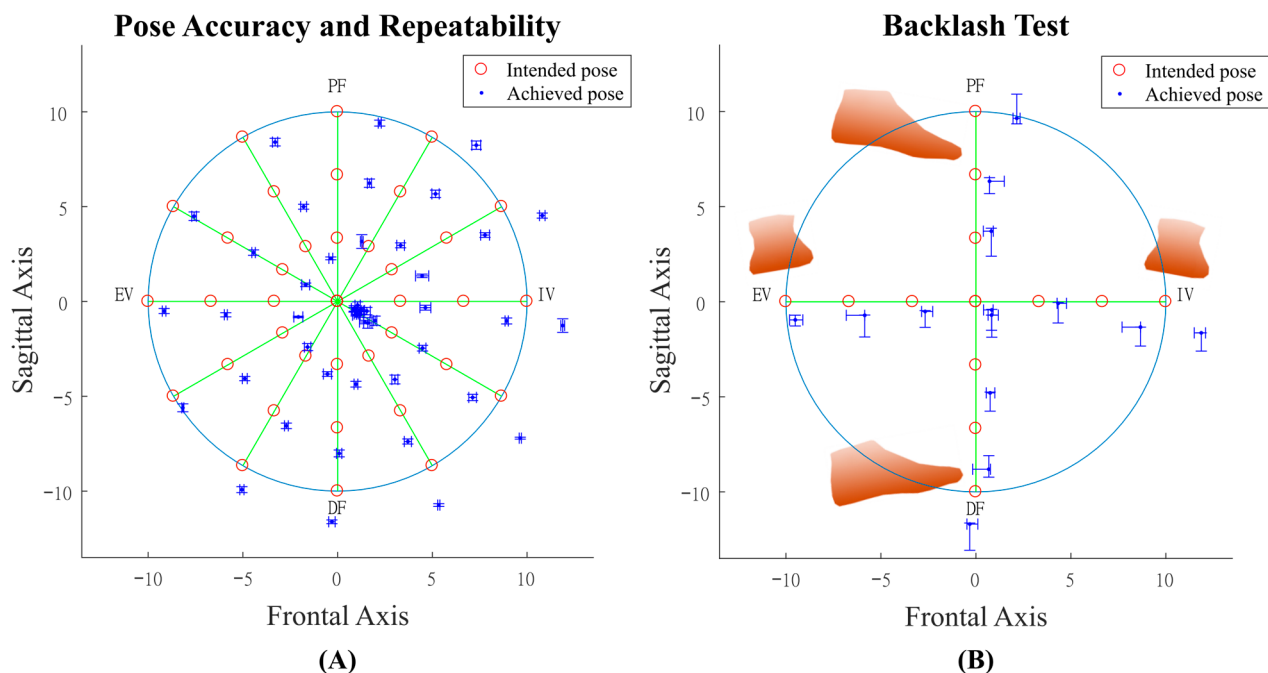


Figure 9. (A) Accuracy and variability test of ankle angles achieved over 240 randomized movements to 48 poses. Radial lines indicate different values for downward direction α . Red circles $\circ$ indicate different values for foot inclination angle θ . Blue dots $\bullet$ indicate the mean of the achieved configuration of the prosthetic foot when targeting each configuration. Standard deviations of foot angle in both sagittal and frontal planes are shown in both tests. Results show error in most configurations including the various “neutral” settings ($\theta = 0$). However, each configuration has high repeatability with little deviation. (B) Ankle backlash test shows that in any configuration, the prosthetic foot is free to deviate slightly from the desired configuration by no more than 1.56 degrees in the sagittal plane and no more than 0.98 degrees in the frontal plane. The configuration with maximum ankle backlash in the PF/DF direction was PF 10 degrees, and in the IV/EV direction, it was EV 6.67 degrees. Inset graphics illustrate the extremes of motion (maximal PF, DF, IV, EV) through sagittal and posterior views, assuming a right foot.

2.5.5. Accuracy Optimization

Initial testing of the TADA’s ankle kinematic performance showed that there must be some inaccuracy in the geometry of the parts or their alignment, as there was imperfect matching of poses achieved to those commanded (see Figure 9A). However, the system’s repeatability allows it to be calibrated to pre-correct for this distortion. Using the results of the initial tests (shown in Figure 9A), we implemented a two-dimensional interpolation to map the intended physical configuration into adjusted commands for direction and inclination based on the initial results. We used *scatteredInterpolant* objects in MATLAB R2021a to perform linear two-dimensional interpolation among scattered (rather than gridded) input data using a triangulation algorithm [35]. We built two interpolating functions—one for α and one for θ —using the observed PF and IV angles as reference coordinates (independent variables; blue dots in Figure 9A) and the original commanded values of α and θ as function values (dependent variables; red circles in Figure 9A). We then used the target PF and IV angles (red circles) to look up adjusted α and θ commands to compensate for the original errors (Figure 10A). Some adjusted commands fell outside the meaningful range; we remedied this by limiting the adjusted θ to a maximum of 10 degrees (2β). The resulting corrections were programmed into the controller to ensure optimal accuracy in later pose commands. Results of this correction are shown in Figure 10B.

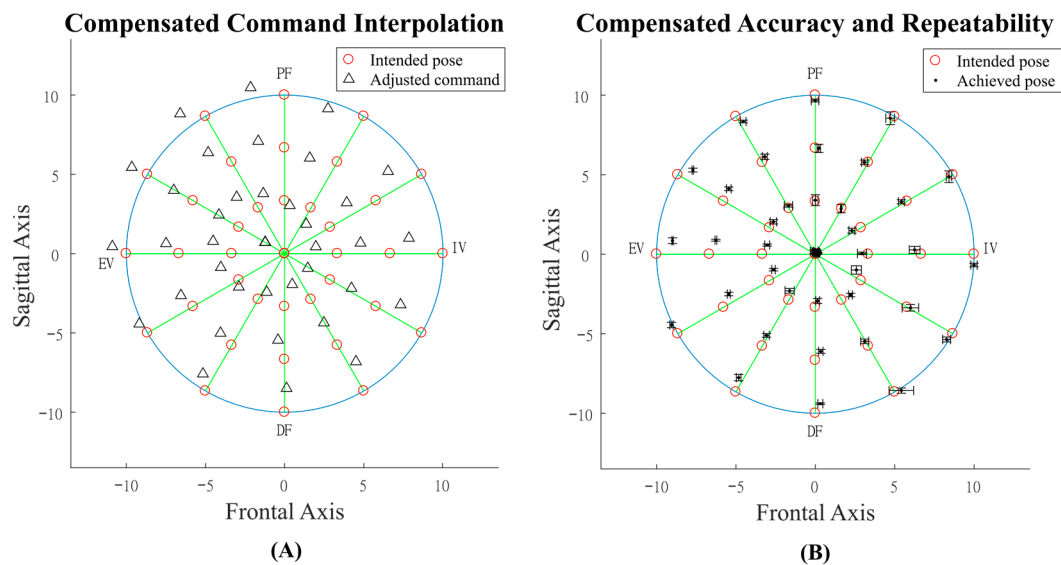


Figure 10. (A) Adjusted commands to improve accuracy, derived from interpolation (MATLAB *scatteredInterpolant*) of raw positioning results displayed in Figure 9A. The black triangles Δ represent compensated commands that should be given to the controller to better zero the device and hence reach desired positions (red circles $\circ$) with reduced orientation errors. (B) These corrections successfully helped to reduce the final positioning error (black dots and standard deviation bars) of the prosthetic ankle over the random 240-configuration test. Part B reprinted with permission from [23].

3. Results

Benchmark and walking trials indicated that the TADA functioned as designed. The mechanism successfully repositioned to all commands in both tests, and held its position until commanded to a new one. No component failure occurred, and no material damage was observed.

Backdriving of the mechanism under body weight in walking was negligible. Figure 7 shows an excerpt of cam movement data from a walking trial with periodic switches between an inclined angle (5 deg EV) and neutral. The wedge cam moves only when actuated by the motor during swing phases, and not when supporting the body-weight loads applied during stance. In total, across nine backdrivability test trials, only 64 samples showed any angle change out of a total of 27,044 samples during stance phases (0.24% of samples), and these few movements were very small and occurred at the beginning of the stance phases when the cam was settling after an actuated movement. The motors did not activate to hold the cams in place during stance.

Figure 8 shows that the ankle can move into any desired position in less than 0.9 s on average (Figure 8A), with movement time proportional to the angle of change for each wedge cam, as expected for motors running at near-constant speed. The regression line predicts movement of 90 degrees in 0.48 s and 180 degrees in 0.83 s. All but one of the movements resulted in wedge cam orientation errors less than 3.2 deg (Figure 8B). This is equivalent to ankle angle error magnitude less than 0.6 deg. The wedge cam orientation errors' frequency density histogram (Figure 8C) showed a skewed distribution with a peak in the 2.5–3 degrees error bin. The distribution was determined to be non-Gaussian using both statistical tests ($p < 0.0005$). Several outliers were present, indicating movements that took unexpectedly long times to reach their target poses (Figure 8A).

Figure 9A shows that angles throughout the target of 10 degrees inclination in any direction are achievable by the TADA mechanism. The foot angles measured by motion capture show errors between the achieved foot orientations (blue dots) and the commands (red circles). The worst mean error was 0.79 ± 0.50 deg in PF/DF and 1.25 ± 0.56 deg in IV/EV. The maximum orientation error magnitude was 3.49 deg (2.28 deg in PF/DF and 2.64 deg in IV/EV), with errors generally biased toward dorsiflexion in the sagittal plane

and toward inversion in the frontal plane. The result angles show low variability in the achieved configurations (PF/DF ± 0.15 deg s.d., IV/EV ± 0.15 deg s.d.).

Figure 9B shows that backlash in the mechanism was modest but present in all configurations. Maximum backlash averaged 1.09 ± 0.29 deg (mean $\pm$ s.d.; max 1.56 deg excess dorsiflexion) in the sagittal plane (PF/DF) and 0.64 ± 0.20 deg (max 0.98 deg excess inversion) in the frontal plane (IV/EV) across all conditions shown. The configuration with maximum ankle backlash in the PD/DF direction was PF 10 degrees, and in the IV/EV direction, it was EV 6.67 degrees.

Figure 10A shows the remapping of intended ankle poses to adjusted commands. Adjusted commands outside the 10-degree circle were projected radially onto it (by limiting θ to 10 deg). The adjusted commands successfully brought the measured configuration into alignment with the intended poses (Figure 10B). The final positioning error when using the adjusted commands was 0.35 ± 0.27 deg in PF/DF and 0.29 ± 0.25 deg in IV/EV, with worst-case errors of 1.23 deg for PF/DF and 1.79 deg for IV/EV. Backlash was not affected by this change.

4. Discussion

The results show that the TADA design is successful in achieving non-backdrivability as intended. The observation of negligible backdriven motion in the wedge cams despite large time-varying external loads experienced in walking (Figure 7) demonstrates that the friction-based wedge cam design is able to support high external loads while successfully isolating the small motors from them. Therefore, the non-backdrivability is a critical contributor to system compactness and weight savings: the TADA uses small actuators because these neither drive nor withstand body-weight forces. Non-backdrivability also provides a fail-safe benefit: it enables the TADA to remain in its most recent pose if it is powered off.

The results also show that the TADA was successful in achieving two-axis motion of the ankle. Movement time (Figure 8) of 0.48 s for a 90-degree cam movement and 0.83 s for 180-degree movement were approximately as predicted from the powertrain design. These movement times indicate the ability to move the ankle from an extreme pose (± 10 deg) to neutral or vice-versa within roughly one swing phase of gait (~ 0.5 s). Therefore, full reversal of ankle angle—such as when turning around on a slope—can be accomplished in two strides. Full turns in walking generally take at least two strides to accomplish, so this movement speed is appropriate for a gradual adaptation strategy, as proposed for a previous semi-active prosthesis [28]. The presence of longer-duration movements—likely due to friction or momentary “sticking”—suggests the need to further optimize the drive train for reliability. A goal for the next revision of this prosthesis is to reduce variability in movement time and achieve all movements in one swing phase—a reduction in movement time that is likely achievable with only modest increases in motor size and quality. Such higher-speed movement would ensure robustness, and if reduced further, would also enable more dynamic adaptations such as lifting the toes during swing phases.

The semi-active design includes several unusual features which proved successful in achieving two-axis control with low system mass. The combination of a non-backdrivable friction-lock wedge cam system with shifted bang-bang control minimizes the power used for actuation. The motors run at full power for short periods of time, and then rest and consume no power at all while the friction lock holds position. Friction computations suggest that the wedge cam face angle could be increased to roughly 10 degrees while retaining this locking property, which may enable further increases in range-of-motion (theoretically exceeding ± 20 deg). The TADA has a build height of only 50 mm and a mass of 550 g excluding the controller and battery; it may be possible to reduce these specifications further with design optimization. The semi-active design means the additional battery and control electronics can also be relatively small and lightweight.

Angles throughout the achievable range of PF/DF and IV/EV were achieved with high precision and repeatability. But, the absolute positioning accuracy under nominal

commands was not as good as intended (see Figure 9A). Several factors could contribute to this inaccuracy, including different mechanical loads, such as the gravitational moment due to prosthesis weight; misalignment between the universal joint and the wedge cams; or manufacturing or assembly inaccuracy in the wedge cam face angles. The total angular excursion in each direction exceeded the 10 degrees (2β) that should be achievable (Figure 9A). The excess motion suggests that some level of manufacturing inaccuracy contributes, such as a 6 deg face angle β instead of the intended 5 deg angle. The imperfect isolation of PF/DF and IV/EV (Figure 9A) further suggests some misalignment in the nominal pose, which may be a consequence of imperfect zeroing of the wedge cam rotation angles or of limitations in the definition of PF/DF and IV/EV directions during motion capture. The system registers the absolute position at startup by rotating each wedge cam a full revolution and detecting a small embedded magnet with a hall effect magnetic sensor on the housing. The zero position is defined by an offset from the magnet's position, which is set visually by the experimenter. Both detection of the magnet and the visual determination of offset could have introduced error in the final zero position.

Whatever the underlying reasons for the pose error, the simple correction mapping used to adjust commands was successful in reducing mean error by 55% in PF/DF and 76% in IV/EV, and reducing the worst-case error by 46% and 32%, respectively. Future versions will improve the initial accuracy further by building in features such as limit switches for better zeroing and/or absolute position encoders for continuous direct configuration measurement. Pre-correction of any remaining misalignment can be built into the controller following a calibration test similar to that used here.

The kinematic planning algorithm based on the downward direction and inclination angle greatly simplified the control calculations and made them more practical. The simple formulas in (3)–(8) are much easier than comparing the rotation matrix itself against its counterpart based on separate PF/DF and IV/EV angles (see Appendix A). Another practical advantage is apparent in the use case of adapting to the terrain angle: if the ground slope can be measured (e.g., through a pylon camera [36,37] or kinematic sensors [38]), it will likely be measured in a pylon reference frame, which will natively define inclination θ and downward direction α .

In steady-state level walking, the TADA is intended to set ankle angles that are equivalent to a prosthetist's ideal alignment of the prosthesis, and essentially leave the ankle fixed in that position as long as the motion continues. In this way, the user can exploit the mechanics of the foot module attached below the TADA as intended by its manufacturers. Within each stride, the load path through the TADA changes from initial contact at the heel to final contact at the toe. Early in ground contact, the ground reaction force acts to force the ankle into plantarflexion, and this external force is counteracted by compressive contact forces at the posterior edge of the wedge cams. Through mid-stance, the ground reaction force advances toward the toe while also shifting mediolaterally under the foot; at each instant, a different point around the circumference of the wedge cam mechanism supports the external ankle moment. Finally, in late contact to toe-off, the mechanism experiences loads similar to the worst-case scenario analyzed above, with the wedge cams supporting the highest loads at their anterior edge. Throughout the movement, the wedge cams never move because the TADA's non-backdrivable mechanism enables "set-it-and-forget-it" use in this case; the system consumes minimal power and requires no active control (only quiescent sensing and state monitoring).

Beyond steady-state level walking, the availability of two-axis controlled movement enables several interesting use cases that may yield biomechanical benefits. The most obvious is adaptation to terrain: the TADA module could be integrated with sensing such as an ankle load cell [39] to detect and match the slope of the ground. Once the slope is detected, local movements like turns or repeated paths could be tracked in real-time with an embedded inertial sensor [27,28,40,41] and used to preemptively adapt the TADA to the known ground slope under the upcoming footfall. Another application is to augment balance during locomotion, such as enhancing lateral balance [1] and steering [42]. In

this usage, the TADA could sense changes in foot placement or movement direction and preemptively move to augment them. A third use case could be perturbation training, in which the TADA would create small disturbances to the ankle angle to lightly disturb the user [43]. Practicing with this mode could train a prosthesis user to be more stable on uneven terrain even without the TADA's assistance. Finally, the TADA could be used to deliberately make changes to the ground contact conditions in order to influence a person's movement. For example, a DF perturbation could facilitate acceleration [44] or an IV/EV perturbation could facilitate a turn. These functions could eventually be coupled with a brain-machine interface to enable feed-forward control of the prosthesis.

A consideration for translating the TADA to commercialization is how to scale it for larger or smaller users. As a standalone ankle module, the TADA can be used with prosthetic feet of different sizes and, therefore, it need not be scaled finely for small increments. However, a few sizes may be envisioned, such as small/pediatric, medium, and large. The key mechanical design parameters must be chosen to respect stress limits on the key components—the mating surfaces of the wedge cams and the pins in the universal joint. The worst-case stress on these parts is driven by the multiaxial ankle moment, which depends on body mass m and foot length l (assumed proportional to body height h). Assuming a constant body mass index (m/h^2), foot length scales with $m^{1/2}$ and therefore ankle moment scales with $m^{3/2}$. For constant stress, the stress-bearing area of the critical pieces must remain proportional to this load such that area also scales with $m^{3/2}$. A proportional scale-up of the TADA's geometry then yields length dimensions scaled by $m^{3/4}$ (square root of area) and device mass scaled by $m^{9/4}$ (cube of length). Thus, a small/pediatric TADA with a 70 kg weight limit is expected to have a mass of 290 g and a build height of 38 mm, and a large TADA with a 125 kg weight limit is expected to have a mass of 910 g and a height of 63 mm. These rough estimates can be used to gauge whether simple scaling is adequate, or whether a redesign for users of different sizes might be necessary. It should be noted that an “oversized” TADA will still function for a smaller user if it can be fitted to the body; there is no lower limit to the loads that work with the mechanism, only an upper limit for strength.

Finally, the wedge cam drive mechanism may not be limited to use in two-axis ankle motion, but could be used in other multi-axis robotic concepts. Unactuated versions of the wedge cam design are prevalent in ductwork to allow rigid pipes to articulate to variable angles; the actuated version could be used for repositionable tentacles or support structures, perhaps with higher wedge angles in cases that allow for higher friction. Alternatively, the semi-active mechanism described here could be converted into a fully-powered mechanism for multiple applications if the friction interfaces in the wedge cam stack were instead supported by bearings and the system was powered by stronger motors.

Limitations

The main limitation of the TADA concept is its design as an inherently semi-active device. Without the ability to articulate under body-weight loads, responsive control of stance phases cannot be achieved. This limitation prevents the TADA from mimicking some features of natural ankle control, such as responsive control of ankle inversion/eversion moment during a stance phase in response to a perturbation [7,8]. Such a response would require an active prosthesis [11,12]. However, the compactness, low mass and low power consumption of the semi-active mechanism stand as the benefits gleaned from trading away this function, while still gaining the ability to adapt to two-axis slopes. Future brain-machine interfaces and pylon-embedded sensors may narrow this performance gap by improving the predictive capabilities of controllers for semi-active mechanisms like the TADA.

One challenge for application of the TADA is the height of the prosthetic foot attached beneath it. The final height of the ankle above the ground is important for the kinematics of foot movement: an ankle near ground level accomplishes nearly pure rotation of the plantar surface, whereas an ankle much higher also causes this surface to translate substantially as

a result of the rotation. In the current assembly, the TADA is attached on top of a standard-height foot, so the TADA ankle center is at roughly the same height as a natural ankle. Therefore, we expect the effects of articulation to match closely the natural ankle's effects (see Figure 1). However, we expect the TADA to be most beneficial when the motions are closest to pure rotation, i.e., when the ankle is mounted on a very low-profile prosthesis. This case would allow for intentional control of the ground contact angle without large shifts in the location of the plantar surface. A future design challenge may be to design a foot prosthesis to achieve such a low center of rotation by incorporating TADA-like mechanisms inside the foot module.

The greatest limitation of the mechanism design itself is the backlash or “slop” in the angle setting. In the current realization, this backlash has a magnitude of roughly one degree, meaning that the ankle can move this amount even when the mechanism is holding a configuration. This backlash is a consequence of the formally over-constrained mechanism: perfect operation requires the centers of the wedge cam faces to coincide with the rotational center of the internal two-axis universal joint. Due to manufacturing tolerances, this can never be perfectly achieved, so the mechanism must be built with a little room for error. In the current device, this backlash is minimized by adding thin shims during installation of the universal joint to ensure the mechanism does not bind. However, the wedge cams incorporate PEEK plastic inserts as the interface surfaces, and the tolerance and dimensional stability of these parts may be inadequate. In future revisions, we will consider replacing the plastic components in favor of metal-on-metal interfaces with tighter tolerances, and we will design a more convenient adjustment mechanism to perform fine alignment of the joints to minimize this backlash.

The imperfect kinematic performance under nominal commands, and, therefore, the need to calibrate and adjust the commands, is an inconvenience that should be improved. As discussed above, it could arise from several imperfections including manufacturing tolerances, initial alignment of the wedge cams, and even axis definitions in the motion capture system. A critical improvement for future versions will be to tighten geometric tolerances and eliminate the alignment steps, to greatly reduce uncertainty about the movement directions. Another improvement would be to incorporate precise absolute angle measurement for the wedge cams, in place of the initial alignment procedure and incremental tracking currently used.

Another challenge is the balance of friction vs. motor power. The semi-active design concept aims to achieve light weight, which promotes smaller motors, but the presence of an intentional friction pathway resists all motion and demands larger motors to overcome it. Substantial effort during tuning was dedicated to achieving a feasible motor size and power settings. Certain configurations are the most taxing; specifically, any move from plantarflexion to dorsiflexion requires the motors to overcome both friction and the gravitational moment of the prosthesis' weight. This problem was ultimately overcome with a slight increase in motor voltage. Future revisions will use better motors and optimize the choice of battery and control electronics to further guard against these problems. These changes are also expected to improve actuation speed and reduce noise.

Because friction plays such an important role in this mechanism, it is important to consider how stably it can be defined. As designed, the TADA requires a friction coefficient of at least 0.05 to ensure non-backdrivability in the wedge cams, and the upper limit interacts with the motor power requirements as described above. The current design was chosen because the PEEK plastic contact has relatively well-characterized friction properties within this range ($\mu = 0.1$) without any need for lubrication. In future work, one goal is to change this material for one with better dimensional stability and tolerance, but this again leads to the challenge of selecting materials and potentially lubricant. In general, it seems likely that either lubricant-free operation or the use of a solid lubricant or lubricant-impregnated solid material would be desirable, to prevent any need for adding lubricant through maintenance. Careful shielding and additional testing in harsh use cases

such as dusty or wet conditions will be necessary to verify that this critical property is preserved in any future mechanism.

Finally, the packaging and form factor of the TADA limit its direct conversion to a final version. Revisions to the housing and component layout are needed for improved compactness and robustness. The protective housings should be nested and enclosed with a rubber seal, the electronics and batteries should be embedded, and the motors should be switched to right-angle drive or offset behind the tibial pylon to reduce height and fit within a standard shoe. The Raspberry Pi 3B was chosen for its ease of implementation as opposed to an embedded controller. It includes a GUI, is flexible, and is very versatile in its functionalities. Some of the limitations of using this single board computer include latency due to use of Python programming and presence of an operating system, and high power consumption. This can be observed by the fact that the highest attainable sampling frequency was 84 Hz. In future work, an embedded controller would be more suitable since it is more compact in size and offers better energy efficiency. With such a controller, real-time reconstruction could be utilized. These and other improvements will improve the practicality of future versions of the TADA.

5. Conclusions

The non-backdrivable, semi-active wedge cam mechanism of the Two-Axis ‘Daptable Ankle (TADA) successfully and repeatably achieves control of the ankle angle in two directions with low weight and power consumption. Orientation control in this mechanism has a simple and intuitive form for easy implementation. Shifted bang–bang motor control achieves maximal-speed movement with no-power rest conditions and adequate accuracy. Some backlash is necessary to prevent binding in the mechanism, but this can be improved with better machining tolerances. Upcoming work will include application testing to determine the effects of two-axis ankle control on walking in persons with amputation, including adaptation to slopes and speeds, augmentation of balance and steering, and perturbation training to promote robust balance.

Supplementary Materials: The following supporting information can be downloaded at: <https://www.mdpi.com/article/10.3390/prosthesis6030049/s1>, Figure S1: Kinematic Relationships; Figure S2: Supplemental Views of TADA.

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Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki, and approved by the Institutional Review Board (or Ethics Committee) of The University of Wisconsin (protocol 2017-0678, approved 8 January 2018).

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Conflicts of Interest: Authors P.G.A. and I.F.E.S. are inventors on a patent covering the technology described in this manuscript. Author P.G.A. was a part owner of Intelligent Prosthetic Systems, LLC. The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Appendix A

Below are the rotation matrices used to represent the movements of the TADA prosthetic ankle (see Ankle Design section). Each matrix converts a column vector expressed in the frame indicated by the lower left subscript into its equivalent vector expressed in the frame indicated by the upper left superscript. Each step of the manipulation is described following the relevant matrix:

$${}^0_1R = \begin{bmatrix} \cos(q_1) & -\sin(q_1) & 0 \\ \sin(q_1) & \cos(q_1) & 0 \\ 0 & 0 & 1 \end{bmatrix}, \quad (A1)$$

rotation of upper wedge cam, controlled by motor 1.

$${}^1_2R = \begin{bmatrix} \cos(q_2) & 0 & \sin(q_2) \\ 0 & 1 & 0 \\ -\sin(q_2) & 0 & \cos(q_2) \end{bmatrix}, \quad (A2)$$

inclination of the upper wedge cam face; $q_2 = \beta = \frac{\pi}{36}$.

$${}^2_3R = \begin{bmatrix} \cos(q_3) & -\sin(q_3) & 0 \\ \sin(q_3) & \cos(q_3) & 0 \\ 0 & 0 & 1 \end{bmatrix}, \quad (A3)$$

rotation of lower wedge cam relative to upper wedge cam (conceptual only; it is a consequence of q_1 and q_5 in (1)).

$${}^3_4R = \begin{bmatrix} \cos(q_4) & 0 & \sin(q_4) \\ 0 & 1 & 0 \\ -\sin(q_4) & 0 & \cos(q_4) \end{bmatrix}, \quad (A4)$$

inclination of the lower wedge cam face; $q_4 = \beta = \frac{\pi}{36}$.

$${}^4_5R = \begin{bmatrix} \sin(q_5) & -\sin(q_5) & 0 \\ \sin(q_5) & \cos(q_5) & 0 \\ 0 & 0 & 1 \end{bmatrix}, \quad (A5)$$

theoretical rotation of the foot back into a toes-forward orientation; controlled in practice through rotation of the lower wedge cam, controlled by motor 2.

The composition of sequential rotations makes the final rotation matrix 0_5R expressing foot-fixed reference frame 5 in coordinates of leg-fixed reference frame 0:

$${}^0_5R = {}^0_1R {}^1_2R {}^2_3R {}^3_4R {}^4_5R = \begin{bmatrix} a & b & c \\ d & e & f \\ g & h & i \end{bmatrix}. \quad (A6)$$

Individual elements of 0_5R are:

$$\begin{aligned} a &= -\cos(q_5)\#5 - \sin(q_5)\#3 \\ b &= \sin(q_5)\#5 - \cos(q_5)\#3 \\ c &= \cos(q_1)\cos(q_4)\sin(q_2) - \sin(q_4)\#7 \\ d &= \cos(q_5)\#4 + \sin(q_5)\#2 \\ e &= \cos(q_5)\#2 - \sin(q_5)\#4 \\ f &= \sin(q_4)\#6 + \cos(q_4)\sin(q_1)\sin(q_2) \\ g &= \sin(q_2)\sin(q_3)\sin(q_5) - \cos(q_5)\#1 \\ h &= \sin(q_5)\#1 + \cos(q_5)\sin(q_2)\sin(q_3) \\ i &= \cos(q_2)\cos(q_4) - \cos(q_3)\sin(q_2)\sin(q_4) \end{aligned} \quad (A7)$$

The numbered arguments in these expressions are:

$$\begin{aligned}
 \#1 &= \cos(q_2) \sin(q_4) + \cos(q_3) \cos(q_4) \sin(q_2) \\
 \#2 &= \cos(q_1) \cos(q_3) - \cos(q_2) \sin(q_1) \sin(q_3) \\
 \#3 &= \cos(q_3) \sin(q_1) + \cos(q_1) \cos(q_2) \sin(q_3) \\
 \#4 &= \cos(q_4) \#6 - \sin(q_1) \sin(q_2) \sin(q_4) \\
 \#5 &= \cos(q_4) \#7 + \cos(q_1) \sin(q_2) \sin(q_4) \\
 \#6 &= \cos(q_1) \sin(q_3) + \cos(q_2) \cos(q_3) \sin(q_1) \\
 \#7 &= \sin(q_1) \sin(q_3) - \cos(q_1) \cos(q_2) \cos(q_3)
 \end{aligned} \tag{A8}$$

The matrix 0_5R is equivalent to a rotation matrix specified directly by a pitch–roll or roll–pitch Euler angle sequence of body-fixed rotations. For a pitch–roll sequence (first plantarflexion θ_{PF} about a leftward-pointing mediolateral y -axis, then inversion θ_{IV} (assuming the right foot) about a forward-pointing anteroposterior x -axis), the matrix is derived in two steps:

$$R_{PF} = \begin{bmatrix} \cos(\theta_{PF}) & 0 & \sin(\theta_{PF}) \\ 0 & 1 & 0 \\ -\sin(\theta_{PF}) & 0 & \cos(\theta_{PF}) \end{bmatrix} \tag{A9}$$

$$R_{IV} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\theta_{IV}) & -\sin(\theta_{IV}) \\ 0 & \sin(\theta_{IV}) & \cos(\theta_{IV}) \end{bmatrix} \tag{A10}$$

$$R_{PF_IV} = R_{PF}R_{IV} = {}^0_5R = \begin{bmatrix} \cos(\theta_{PF}) & \sin(\theta_{IV}) \sin(\theta_{PF}) & \cos(\theta_{IV}) \sin(\theta_{PF}) \\ 0 & \cos(\theta_{IV}) & -\sin(\theta_{IV}) \\ -\sin(\theta_{PF}) & \sin(\theta_{IV}) \cos(\theta_{PF}) & \cos(\theta_{IV}) \cos(\theta_{PF}) \end{bmatrix}. \tag{A11}$$

Individual rotations can be solved for numerically to obtain q_1 , q_3 , and q_5 that match the 0_5R and R_{PF_IV} matrices. A similar matrix can be derived from a roll–pitch rotation sequence or using pylon-fixed rotation axes. In any of these cases, derivation through Euler angles results in some distortion of the movement relative to the intended direction-and-angle control (e.g., see the zero value in R_{PF_IV} that may not be zero in 0_5R). This occurs because two sequential pitch–roll or roll–pitch rotations are incomplete for describing a three-dimensional rotation as implemented by the TADA; a third rotation such as yaw would be necessary. The degree of distortion can be evaluated by including that third rotation and determining its required value, similar to the error evaluation associated with Equation (8) above. Practically, the conclusion is that a simple two-rotation Euler sequence is not recommended.

The matrix 0_5R is also equivalent to a rotation matrix specified directly by the (θ, α) pair, which can be transformed into an axis–angle formulation. For a rotation of θ about unit axis $\hat{\omega} = [\omega_x, \omega_y, \omega_z]$, a rotation matrix can be built using Rodrigues' formula [45]:

$$R_{ax_ang} = \begin{bmatrix} C + \omega_x^2 \bar{C} & \omega_x \omega_y \bar{C} - \omega_z S & \omega_x \omega_z \bar{C} + \omega_y S \\ \omega_x \omega_y \bar{C} + \omega_z S & C + \omega_y^2 \bar{C} & \omega_y \omega_z \bar{C} - \omega_x S \\ \omega_x \omega_z \bar{C} - \omega_y S & \omega_y \omega_z \bar{C} + \omega_x S & C + \omega_z^2 \bar{C} \end{bmatrix}. \tag{A12}$$

where $C = \cos \theta$, $\bar{C} = (1 - \cos \theta)$ and $S = \sin \theta$ for compactness. The rotation axis $\hat{\omega}$ is in the plane orthogonal to the TADA's upper surface; $\hat{\omega}$ is rotated 90 deg relative to the direction vector α ; and $\omega_z = 0$. Individual rotations can be solved for by numerical solution for q_1 , q_3 , q_5 that match the 0_5R and R_{ax_ang} matrices.

Appendix B

This section provides an analysis of the backdriving rotational moment and friction holding moment in the wedge cam mechanism. For simplicity, we assume Coulomb friction and a point–load contact model. Forces are localized at the contact point farthest away from the joint center toward the applied load, which is in fact the location of contact in

the real mechanism due to backlash in the wedge cams. The result is identical to using a distributed load model (Figure 5). Figure A1 shows the upper wedge cam under loading at the forefoot (see Figure 5).

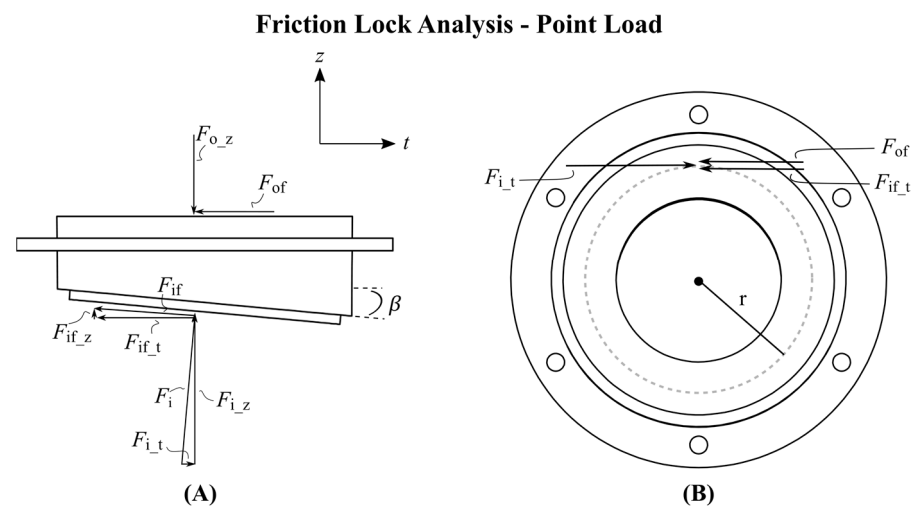


Figure A1. Diagram of applied static loads and friction holding forces to prevent backdriving in the wedge cam mechanism: (A) front view (toes pointing out of the page); (B) bottom view (toes pointing up). The analysis assumes localized loading at the point farthest from the joint axis. The result is equivalent to the distributed load case (Figure 4).

The causal load is applied to the inclined face in the z -direction, F_{i_z} , to support the ankle moment M_{ankle} (see Figure 5):

$$F_{i_z} = M_{\text{ankle}} / r. \quad (\text{A13})$$

This applied load elicits a reaction force normal to the inclined surface, F_i , which also has a component in a direction tangential to the circumference of the wedge cam, F_{i_t} , due to the surface inclination:

$$F_{i_z} = F_i \cos(\beta), \quad (\text{A14})$$

$$F_{i_t} = F_i \sin(\beta) = F_{i_z} \tan(\beta). \quad (\text{A15})$$

The tangential component of the normal force, F_{i_t} , tends to backdrive the wedge cam with moment M_{back} due to its action at a distance r from the rotation axis (related to (9)):

$$M_{\text{back}} = r F_{i_t}. \quad (\text{A16})$$

This rotational moment is opposed by friction on both the top and bottom surfaces of the wedge cam. The normal reaction force F_i establishes the capacity for a friction force on the inclined surface, F_{if_cap} , approximated by a coefficient of Coulomb friction, μ . To analyze the holding limit, we assume incipient slip, such that the actual friction force F_{if} equals this limit:

$$F_{if} = F_{if_cap} = \mu F_i. \quad (\text{A17})$$

The friction force capacity also has components in the z and t directions, F_{if_z} and F_{if_t} :

$$F_{if_t} = F_{if} \cos(\beta) = \mu F_i \cos(\beta). \quad (\text{A18})$$

$$F_{if_z} = F_{if} \sin(\beta) = \mu F_i \sin(\beta). \quad (\text{A19})$$

On the upper, orthogonal face of the wedge cam, a normal force F_{o_z} opposes all the z -direction forces on the inclined surface, and establishes a similar friction force:

$$F_{o_z} = F_{i_z} + F_{if_z} \quad (A20)$$

$$F_{of} = \mu F_{o_z}. \quad (A21)$$

All the tangential friction forces act with moment arm r about the rotational axis. To achieve static equilibrium, the holding moment from the friction forces must balance the backdriving moment from the normal force:

$$r(F_{if_t} + F_{of}) = M_{back}. \quad (A22)$$

Substituting (A13) through (A21) into (A22) yields an estimate of the friction coefficient necessary to prevent backdriving of a wedge cam with a given angle β :

$$\mu \geq \frac{1 - \cos(\beta)}{\sin(\beta)}. \quad (A23)$$

Rearrangement of (A23) also yields an estimate of the maximum wedge cam angle β that can be held by a material interface with a given friction coefficient μ (identical to (10)):

$$\tan(\beta) \leq \frac{2\mu}{1 - \mu^2}. \quad (A24)$$

For a desired wedge cam angle $\beta = 5$ degrees, Equation (A23) gives a required friction coefficient $\mu \geq 0.044$. For a designed friction coefficient of 0.1 based on the materials at the interface (PEEK plastic and aluminum), the maximum wedge cam angle estimated from Equation (A24) is 11.42 degrees. By both calculations, the estimated safety factor of the TADA as designed is roughly 2.25.

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