

Can a Biomimetic Ankle Joint with Proprioception Provide Feedback On Its Operating Environment?

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Abstract

In current prosthetic devices permitting rotation, one-degree-of-freedom pin-joint-based ankle joints are primarily used, while two-degree-of-freedom devices remain in research; however, it has been shown that they improve the balance of the user compared to one degree of freedom. These research-based devices also predominantly make use of pin joints, which are inherently less kinematically accurate than rolling contact joints of the human anatomy. A two-degree-of-freedom rolling contact joint is multiplanar, and, therefore, rigid links to constrain the joint cannot be used, as this could only operate in one plane. Using soft multiplanar links analogous to ligaments would not only constrain the joint but also allow for characterisation of the joints by measuring the displacement of the links, which can be used to estimate joint angle for traditional control. A simplified version of the tibiotalar and subtalar joints of the ankle was made, forming a joint complex capable of motion in the coronal and sagittal planes. Silicone links were produced with a silicone-based infrared waveguide embedded to provide proprioception of the joint. The anterior talofibular, calcaneofibular, tibiocalcaneal, and superficial tibiotalar ligaments of the ankle were manufactured as deformable waveguides and tested at 10+ % strain to verify linearity. The R^2 value of all the ligaments when compared with a linear regression model was above 0.996 between the operating range of 0-6 % strain. Then the joint complex as a whole was manipulated in the sagittal and coronal planes, providing evidence that there were unique behaviours in the ligament stretch demonstrated in each discrete plantar position, dorsiflexion, plantarflexion, inversion, and eversion. This could then be used in a control scheme for the joint to identify the planar position of the joint, which could then be used to drive and sense two planes of a prosthetic foot. This would ultimately improve balance and stability by permitting the foot to adapt to the floor in both planes with anatomical-like compliance, while allowing for the prediction of the geometry for a control scheme with active control for balance and stability.

Author's declaration

I declare that this thesis is a presentation of original work and that I am the sole author. This work has not previously been presented for a degree or other qualification at this University or elsewhere. All sources are acknowledged as references.

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1 Introduction

The simplification of human joints in prosthetics is a sound engineering decision; it enables design under constraint to fabricate an electromechanical replacement for an otherwise exceedingly complex biological system. However, this mechanical simplification could ultimately be hindering the design of the next generation of prosthetics. As current work is focused on optimising a mechanical system built on a simplification, the mechanism has a hard limit, falling short of achieving the motion of the original system. Work done by Russell implementing this methodology produced a mechanical knee joint capable of replicating the motion of the human knee in the sagittal plane [37]. Russell replaced the pin joint with a rolling contact joint, matched to the geometry of the human knee condyles, constrained by rigid sprung ligaments. Feedback was provided by measuring the linear extension of the springs in the ligaments, encoding joint position[37].

The anatomical ankle permits 6 degrees of freedom from the tibia to the ground plane, where the rotational degrees of freedom allow the joint to comply with non-flat surfaces [2, 1]. The 3 translational degrees of freedom indicate an additional level of compliance, specifically in the position of the joint axes, that could improve the ability of the foot to match the ground plane and increase balance and stability [2]. This project aims to create a biologically inspired ankle joint permitting multiple degrees of freedom by using anatomical geometries of the joint surfaces. To preserve anatomical motion, the joint will be constrained by multiplanar links with motion feedback being provided by measuring the displacement of the links. Furthermore, the project aims to test the validity of this design in both producing anatomical-like motion, as well as giving sufficient sensor feedback to estimate the joint angle of the devices. This will form the foundation of an artificial ankle joint that has similar compliance to the ground surface as the anatomical ankle while providing positional feedback to actuate the degrees of freedom and control of balance. The aims and objectives to achieve this have been outlined in section 3.

2 Literature Review

The following literature review provides an insight into the anatomy of the human ankle joint and how it behaves during normal walking, clinically referred to as gait. As an extension of this, a review was conducted of the current methods of foot-ankle prosthetics in order to gain insight into the direction of the joint design. Potential methods of sensing are then explored, and how different sensing methods could aim to mimic the human anatomy.

2.1 Prosthetic Users

In the UK alone, an estimated 11500 lower limb amputations occur each year [16]. Of this cohort, the average age is 70 years old, with men disproportionately outweighing women 3:1 [16]. It has been reported that an estimated 7000 people undergoing amputation have diabetes, predominantly due to increased risk of peripheral vascular disease [16]. However, traumatic amputations are the cause of 10% of all amputations, with 3% due to malignancy, and a further 1.8% due to congenital limb deficiency. Between the years 2003 and 2013, it was reported that of all the lower limb amputations, 22,645 were transfemoral, or above-knee, and 19,658 were transtibial, or below-knee [16].

2.2 The Joints of the Ankle

The foot is a complex biological mechanism, containing a large number of bones, ligamentous tissue and fascia. Understanding the local anatomy and structures is key to gaining insight into the function and roles of the bone surfaces and ligaments in a kinetic and kinematic context.

2.2.1 The Tibiotalar Joint

The tibiotalar joint shown in Fig. 1 is often referred to as a tenon and mortise joint, as the tenon, like the trochlea of the talus bone, fits into the mortise formed by the distal

end of the tibia and fibula [5, 31]. This geometry is considered to contribute to the high stability of the joint [5]. The tibiotalar interface provides the load-bearing for the joint, while the distal fibula does not provide any direct load bearing; its structure is key to the function of the joint [5]. A majority of the motion of the tibiotalar joint is in the sagittal plane, acting primarily as a hinge joint; however, there is indication in the joint geometry to suggest it has smaller movements in other planes [5]. The posterior taper of the superior articulating facet of the talus bone indicates that when the posterior portion of the bone is dominating in the joint interface, there may be some added movement in other planes, as it is less closely *packed* than when the anterior portion of the facet is dominating the joint [5, 31].

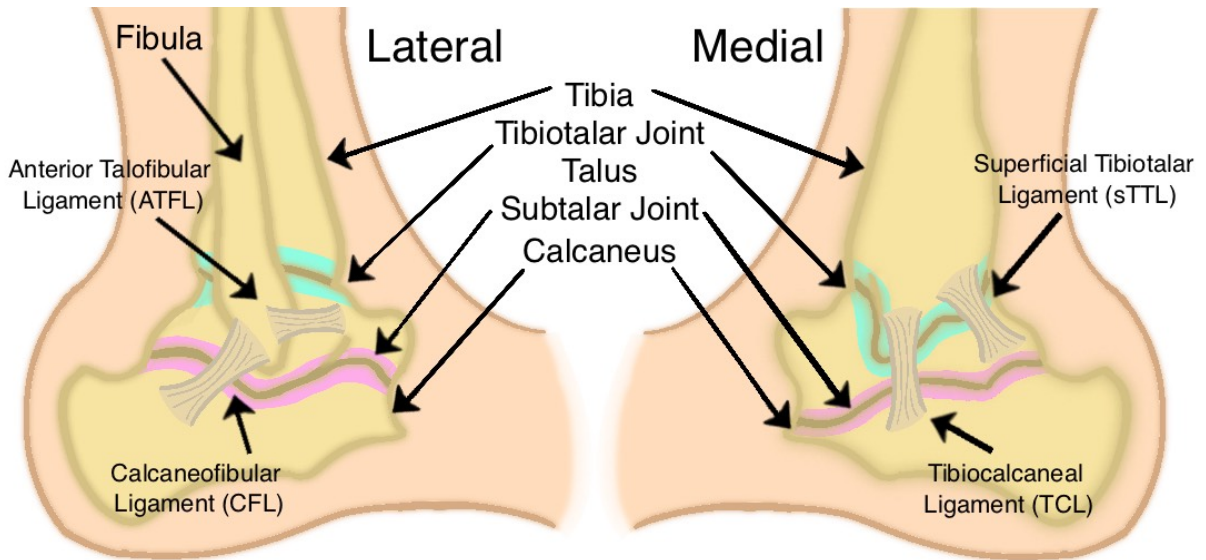


Figure 1: The local anatomy of the hind foot, showing the tibiotalar and subtalar joint complex in the medial and lateral viewpoints. Also shown are the anatomical positions of the lateral anterior talofibular and calcaneofibular ligaments, and the medial superficial tibiotalar and tibiocalcaneal ligaments.

2.2.2 The Inferior Tibiofibular Joint

The inferior tibiofibular joint, or the ankle syndesmosis, is a non-articulating joint connecting the tibia and the fibula at the distal ends, shown in Fig. 2 [31]. Structurally, it is a key part of the tibiotalar joint but can be considered its own distinct joint [5]. The joint is predominantly non-articulating in nature; structurally, its primary role is to form

the stable *mortise* structure out of the malleoli of the distal tibia and fibula, and the dome structure of the tibia [31]. This formation, in conjunction with the matching dome of the talus, forms a very strong joint [31]. The ligaments of the syndesmosis support this structure by resisting motion in all planes [31]. Secondly, the malleoli also provide the origin points for the lateral and medial ligaments on the fibula and the tibia, respectively [31, 5].

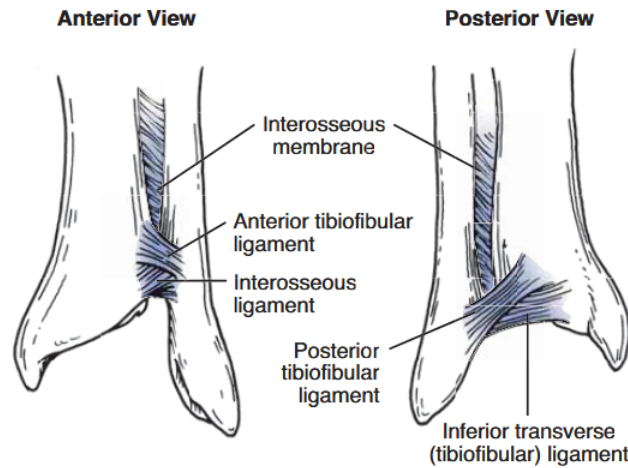


Figure 2: The inferior tibiofibular joint. Taken from [5]

2.2.3 The Subtalar Joint

The subtalar joint is located inferior to the tibiotalar joint, involving the calcaneus and the talus [5]. Shown in Fig. 1, the joint consists of a pair of articulated facets anteriorly and another facet located posteriorly. The anterior pair are convex on the inferior talus and concave on the superior calcaneus; however, the posterior facet is concave on the inferior talus and convex on the superior calcaneus [5]. This geometry allows for stable motion in both the coronal and transverse planes, with the range of motion in both planes being almost equal. [5, 31]. Therefore, in conjunction with the tibiotalar joint, motion is permitted in all three planes [31]. The motion of this joint can be akin to the motion of a helical surface during non-weight-bearing movement; the calcaneus is permitted to translate anteriorly along the rotational axis of the subtalar [31]. This occurs as it rotates clockwise [31]. Since the motion of the joint is in all three planes, it is difficult to isolate

the rotations of the joint clinically [5].

2.3 The Motion of the Ankle and Ligaments

The motion of the foot, and by extension the mechanics of the ankle, is suitably complex. Understanding motion and the roles of individual joints and the associated ligaments is key to isolating the crucial elements that can be used to biomimetically inspire joint design.

2.3.1 Dorsiflexion and Plantarflexion

Dorsiflexion and plantarflexion are defined as the directions of the motion of the foot in the sagittal plane, shown in Fig. 3. Dorsiflexion is the positive movement in the axis of the sagittal plane, bringing the foot to a proximal position. Whereas plantarflexion is the negative movement, bringing the foot to a distal position. The primary contributor to dorsi-plantarflexion of the foot is the tibiotalar joint [5]. The geometry of the articulating facet of the talus in the tibiotalar joint is widest anteriorly [5]. This ultimately means that the joint is more stable when in dorsiflexion than plantarflexion [5]. The tibiotalar joint is supported by a number of soft tissue structures during motion. Medially, the joint is supported by the deltoid ligament or the medial collateral ligaments, a triangular-shaped group of ligaments [5]. Primarily providing resistance to eversion, the deltoid ligament consists of the anterior tibiotalar, posterior tibiotalar, tibionavicular, and tibiocalcaneal ligaments [5]. Laterally, the tibiotalar joint is supported by the lateral collateral ligaments, providing resistance to inversion [5]. It consists of the anterior talofibular, the posterior talofibular, and the calcaneofibular [5]. The inferior tibiofibular joint, as mentioned previously, is non-articulating, connected by the anterior and posterior tibiofibular ligaments and the interosseous membrane [5, 31]. It is the soft tissue connection that keeps the joint rigid; it is this non-articulating nature that leaves it commonly susceptible to injury [5]. The range of motion of this joint is reported to be 10° to 20° in dorsiflexion and 40° to 55° in plantarflexion [31].

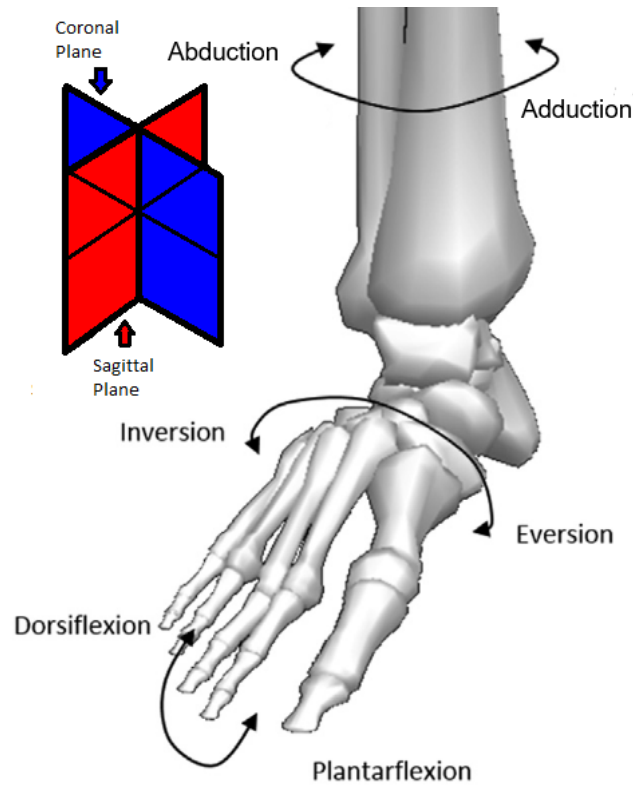


Figure 3: The directions of the movements of the foot, adapted from [5]

2.3.2 Abduction and Adduction

Adduction and abduction are defined as the directions of the motion of the foot in the transverse plane, shown in Fig. 3. Adduction is the positive movement in the transverse plane, bringing the foot to a medial position. Abduction is the negative movement and brings the foot to a lateral position. All of the ankle joints allow motion in multiple planes to a degree; it can be considered that adduction and abduction come from the motion of a combination of the joints [5]. Abduction is permitted by the subtalar and by the transverse tarsal joint as they are part of the same functional unit [5, 31]. This fact makes measuring the specific contribution to ab-adduction difficult, and to the author's knowledge, it has not been reliably isolated [31].

2.3.3 Inversion and Eversion

Inversion and eversion are defined as the directions of the motion of the foot in the coronal plane, shown in Fig. 3. Inversion is the positive movement in the coronal plane, bringing the sole to face the midline. Eversion is the negative movement, bringing the sole to face away from the midline. Inversion and eversion are primarily permitted by the subtalar joint, and secondarily provided by the transverse tarsal joint [5]. Support for the joint is provided by various surrounding soft tissue structures. Most crucially, the interosseous talocalcaneal ligament connects the calcaneus and the talus [5]. The insertion point is on the distal talus, between the articulating facets of the subtalar joint [5]. Likewise, for the calcaneus, the insertion point is on the proximal end, between the facets for the subtalar joint [5]. This arrangement allows for stabilisation and strong resistance to inversion/eversion, while not interfering with the joint facets [5]. The lateral talocalcaneal ligament and the anterior talocalcaneal ligament provide further connection to the joint [5]. However, compared to the support provided by the interosseous talocalcaneal ligament, the talocalcaneal ligaments are weak [5]. Due to the difficulties of isolating the range of motion of the subtalar joint, only an estimate has been inferred by measuring the inversion/eversion of the calcaneus [31]. This was stated to be approximately 22.6° inversion and 12.5° eversion [31].

2.3.4 Supination and Pronation

Combining the motion of all three planes creates what is known as supination and pronation [5]. It is characterised by the direction of the plantar, or the sole [5]. Supination can be defined as a combination of plantarflexion, adduction, and inversion of the foot [5]. As a result of this combined motion, the sole faces medially [5]. For pronation, it is a combination of dorsiflexion, abduction, and eversion of the foot, causing the sole of the foot to face laterally [5].

2.4 Ligament Anatomy

The lateral ankle ligaments, the anterior talofibular ligament (ATFL) and the calcaneofibular ligament (CFL), both have origins on the fibula [27]. For the ATFL, the origin is 10-13.2 mm from the anterior border of the fibula [27]. The origin of the CFL is found 5.4-8.5 mm proximal to the tip of the fibula [27]. The ATFL has been found to have a length of 12-24 mm, and the CFL is 18.5-35.8 mm [27]. The ATFL talar insertion is found 14.2-18.1 mm to the subtalar joint or 11.3-14.8 mm to the anterolateral corner [27]. The CFL calcaneus insertion is found 12.1-13 mm from the peroneal tubercle [27]. The medial ligaments, the superficial tibiotalar ligament (sTTL) and the tibiocalcaneal ligament (TCL), both have origins on the tibia [6].

The tibial origin of the TCL is 6 mm from the distal centre of the intercoliclar groove, and the tibial origin for the sTTL is 3.5 mm from the inferior centre of the intercoliclar groove [6]. The length of the TCL was found to be 25.6 ± 4.5 mm [4]. While the length of the sTTL was found to be 14 ± 3.7 mm [29]. The talar insertion of the sTTL was found to be 10.4 mm anteroposterior to the talar posterolateral tubercle [6]. The calcaneal insertion of the TCL was found to be 8 mm from the sustentaculum tali [6]. The interosseous talocalcaneal ligament (iTCL) had several anatomical variations across different cadavers studied by Jotoku et al, the most common of which was the banded type with n of 37 [22]. The length of the interosseous talocalcaneal ligament (iTCL) was found to be 10.0 ± 3.5 mm [22].

2.5 Gait Analysis

Understanding the static motion of the joint allows an understanding of the motion of the joint from the perspective of minimum motion requirements in a design specification. However, the dynamic motion of a joint is still important to understand, as this is the majority use case of a joint. This understanding leads to adding or subtracting features from a joint that a purely static motion understanding does not give, as well as leading

to insight into what a control scheme for this joint would require.

2.5.1 The Kinematic Chain of the Leg

Gait analysis is the umbrella term for the various analytical methods of study of the leg during walking, or gait [49]. Kinematic gait analysis is the study of the mechanical properties of the leg under motion [49]. While kinetic gait analysis is the measurement of forces occurring in the leg during gait, primary focused on the interaction between the foot and the ground [49]. Human locomotion is an intricate multi-variable kinematic and kinetic problem [49]. Each variable has a direct or indirect effect on the other variables in the system. Therefore, the isolated study of a specific anatomy will give a different result than it would when studied in gait as a whole. Walking is a cyclical motion, following a consistent pattern. This allows a single cycle to be isolated and studied [47]. This is known as the gait cycle, shown in Fig. 4, which is defined as starting when an observed foot strikes the ground, known as the initial contact [47]. This also defines the start of the first major phase of the cycle, known as the stance phase, accounting for 60% of the cycle [47]. During the stance phase, following the initial contact is the loading response, where the observed foot transitions to foot flat at 10% of the cycle [47]. At 30% of the cycle, mid-stance occurs, where the trailing limb has now swung to be aligned with the observed limb [47].

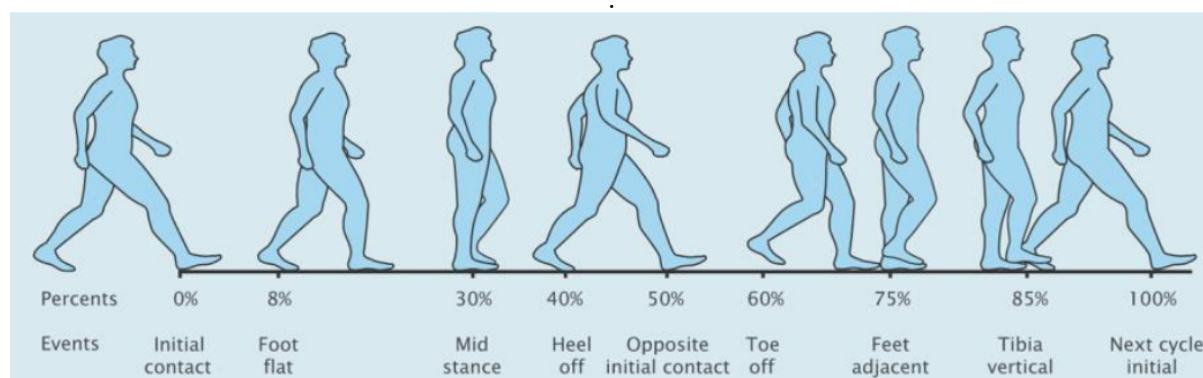


Figure 4: The gait cycle marking the key events at the average percentage progression of occurrence. Adapted from [47].

The 10% to 30% period is the first half of the single limb support for the observed limb

[47]. As the non-observed limb swings ahead, heel off occurs in the observed limb at 40% [47]. At 50%, initial contact occurs in the non-observed limb, starting the loading response and ending the single limb support for the observed limb [47]. The pre-swing is the last 10% of the stance phase, where the body shifts its centre of mass forward, preparing for push-off. This is the transition to the swing phase [47]. At 60%, toe-off occurs for the observed limb while the non-observed limb transitions to foot flat [47]. The swing phase accounts for the remaining 40% of the cycle and is the period of the cycle where the observed limb is no longer in contact with the ground. [47]. Starting at toe-off, the observed limb swings to be in line with the non-observed limb, now in its mid-stance, at 75% of the cycle [47]. Following this, the limb swings ahead of the non-observed limb, preparing for contact with the ground [47]. This is the initial contact of the next cycle and is the end of the swing phase and the cycle as a whole.

2.5.2 Behaviour of the Ankle in Gait

In the context of the foot specifically, the cycle can be further decomposed to understand the motion of the foot in normal gait. For the foot, the stance phase can be decomposed into three distinct sub-phases, shown in Fig. 5, the first of which is the heel rocker phase [5]. The heel rocker phase starts at the beginning of the gait cycle with initial contact [5, 47]. Here, the ankle is marginally plantarflexed, allowing the posterior end of the calcaneus to act as a pivot with the contact surface [5]. During the heel rocker phase, plantarflexion occurs, lowering the foot towards the ground [5]. During foot flat, the ankle rocker phase occurs; this is where the ankle transitions from a state of plantarflexion to dorsiflexion while the foot is in full contact with the ground [5, 47]. This means that the lower limb, or shank, of the observed leg is being permitted to rotate forward by the ankle joint, progressing the body forward [5]. The final sub-phase is the forefoot rocker phase, which is where the foot is permitted to rotate around the forefoot [5]. It starts at heel-off, where the foot again begins plantarflexion, and continues till toe off, when the foot reaches maximum plantarflexion [5, 47]. Once toe-off has been achieved and the

swing phase has started, the foot begins to dorsiflex [5]. This is so the foot is able to avoid contact with the ground while the swing progresses [5].

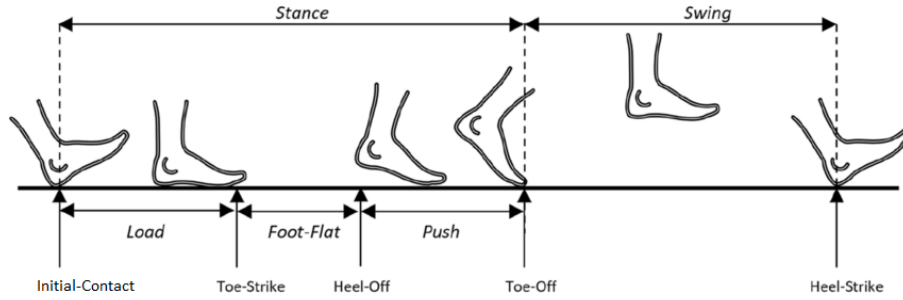


Figure 5: The foot during the gait cycle. Taken from [52]

The dorsiflexion decreases after the midstance, which then reaches a neutral, slightly plantarflexed position, and the next cycle begins [5]. This movement is also joined by motion at the subtalar joint, contributing 15° of inversion to the motion of the foot [5]. The slight plantarflexion of the foot is key to the initial contact and loading response at the start of the cycle. As mentioned previously, the tibiotalar joint is more stable in dorsiflexion, whereas stability in plantarflexion is more reliant on the soft tissue structures [5]. This gives the ankle a certain amount of compliance during plantarflexion, allowing it to behave passively upon initial contact [8]. This allows the ankle to be compliant to the contact surface as it makes contact, allowing the external environment to have a factor in its position [8]. On the other hand, while in dorsiflexion, the foot is far more stable due to the structure of the joint, resisting external factors [5, 8]. Therefore, during toe off and the forefoot rocker phase, when the ankle is generating the force to lift the limb from the contact surface, it applies a force on the contact surface in an attempt to achieve a desired position, where it is the surface that becomes the compliant variable [8].

2.6 Current Foot Prosthetics

Prosthetics is a wide research area with many subsets of design. The two major design types are passive and powered prosthetics. Passive rely primarily on resistance and

stiffness to provide support, whereas powered devices make use of actuators to provide the motion their native muscles would have done, instead of having to compensate with other parts of the body. This provides a starting direction for the initial design of prosthetics, comparing the advantages and disadvantages of existing designs.

2.6.1 Passive Foot Prosthetics

Passive prosthetics are unpowered prosthetic devices that do not provide any active movement. Their function traditionally is to allow for the introduction of basic support and mobility. Recent designs in passive prosthetics aim to increase functionality and introduce features to further assist mobility. Xiu et al aim to deviate from the conventional sagittal plane restricted pin-jointed ankle by replacing it with a novel 2 DOF mechanism [50]. Using the pre-existing Omni-Wrist III mechanism, the authors recognise the similarity in movement with the human ankle, shown in Fig. 6 [50]. The approach by the authors replaces a pin-jointed ankle with the Omni-Wrist III. [50].

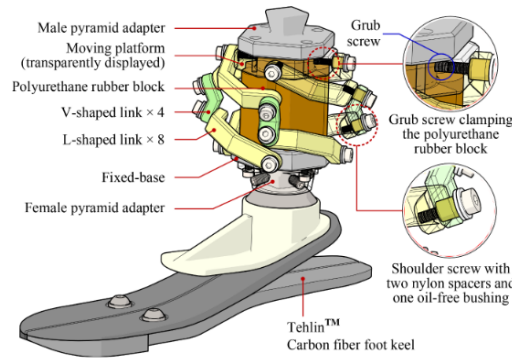


Figure 6: The 2 DOF passive ankle implementation with the Omni-Wrist III designed by Xiu et al. Taken from [50].

The Omni-Wrist II is a parallel plate manipulator, with 4 sets of 4-bar mechanisms, providing orientation of the end plate in two degrees of freedom [50]. The Omni-Wrist III provides little resistance to movement, unlike the ankle; therefore, the authors designed a polyurethane rubber block that would insert inside the mechanism, providing resistance [50]. The rubber insert provided stiffness close to that of an able-bodied human, although it still differs from the able-bodied data; a clear trend is present [50]. The insert consists

of a 30 mm radius cylinder as the main body. In the dorsi-plantarflexion direction, there are two identical 10 mm radius nodes, implying an equal resistance in both directions in the plane [50]. For inversion is a 23.3 mm rectangle protruding 6 mm from the main body of the block; however, for the inversion protrusion, there is a 13 ° taper on the edge of the block [50]. This implies that the design calls for more resistance in the eversion direction than the inversion direction in the plane, due to the added material and therefore increased stiffness.

Piazza et al designed the SoftFoot ankle-foot prosthesis, which returns to a sagittal plane pin joint; however adapts the foot to include further motion, shown in Fig. 7. The methodology for the SoftFoot was to implement the windlass mechanism in a foot prosthesis [32]. The bones of the foot are organised into two primary arches: the longitudinal arch that forms the primary action for the windlass [32]. The windlass mechanism allows the foot to withstand the forces of the human body while being elastic and compliant [32]. The contribution of the windlass mechanism allows the foot to provide adaptability to irregular terrain, balance, and stability [32]. The SoftFoot consists of 5 independent linkage chains that are attached by connected arms, forming a triangle with the ankle joint, matching the geometry of the plantar structures and the windlass mechanism [32].



Figure 7: The SoftFoot designed by Piazza et al. Taken from [32].

Each linkage chain consists of 10 links attached to each other elastically, with each link hosting a pulley structure to allow for an analogue tendon to run through the chain [32]. A linkage at the back of the foot, mimicking the calcaneus, connects all the chains. This linkage also connects the chains to the posterior bar, which is connected to the ankle joint on the other end [32]. Here, a spring also connects each final linkage to the posterior bar, which can store energy to be deployed at push-off [32]. This design allows the prosthesis to be compliant with the shape of uneven terrains and surfaces, while maintaining a robust and passive design [32]. Gait data has yet to be published at the time of writing, but the initial surface testing of the device has shown versatility in the ability to adapt to various surfaces in the sagittal plane, which would increase balance and stability in a prosthesis [32].

To the knowledge of the author, the only example of a foot ankle prosthesis using human bone geometry is a passive prosthesis designed by Jin et al, shown in Fig. 8 [21]. Jin et al used a patient-specific CT scan of the healthy leg of a transtibial amputee [21]. They segmented and smoothed the bones of the tibia and fibula (fused as the shank), talus and

calcaneus [21]. The smoothed bones were mirrored to form an ankle prosthesis for the amputated leg. The joint was constrained by a urethane casing surrounding the ankle joint with voids to accept the bones [21]. Finite element analysis of the system indicated that the joint had good variable stiffness and was capable of anatomical-like motion [21]. During gait testing of the device with the patient, the rotation in the anatomical planes was measured for both the prosthesis and the healthy ankle at 3 different speeds and compared directly [21]. The joint angle was able to match the trend of the healthy ankle in all three planes. However, it was noted that the peak moment values were generally higher and shorter compared to the healthy ankle [21]. This was attributed to the material design of the urethane sheath [21].

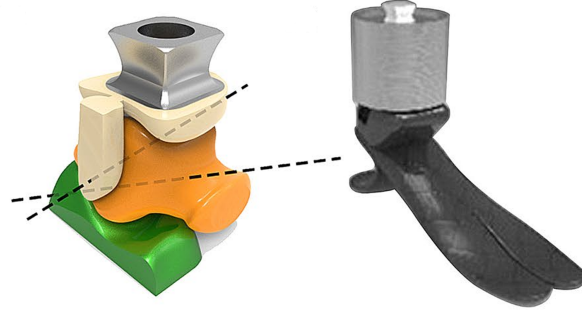


Figure 8: The smoothed bone geometry of the shank, talus, and calcaneus used by Jin et al to implement in their passive prosthesis. Taken from [21].

2.6.2 Powered Ankle-Foot Prosthetics

Passive prosthetics provide no active movement; this is a power gap that the residual biological limb is left to compensate for, causing asymmetries in gait [13]. Powered prosthetics aim to provide this power by incorporating active components. The *Utah Leg*, shown in Fig. 6, developed by Tran et al is a transfemoral powered prosthesis implementing a semi-active knee and a powered ankle joint [46]. The semi-active knee joint is an iteration upon the author’s previous works, a novel approach that uses a bar linkage which can vary the leverage it imparts on the joint by changing its distance from the centre of rotation [46, 45, 24].

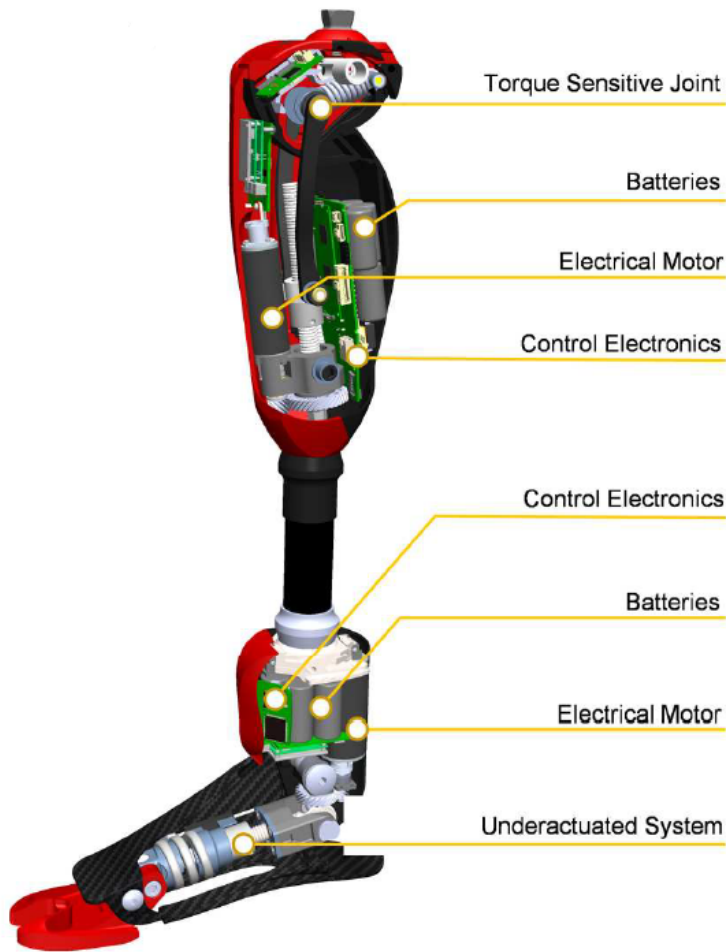


Figure 9: The Utah Leg, designed by Tran et al, showing the internal mechanism. Taken from [45].

The proposed ankle joint actually provides active motion for both the toe and the ankle in the same mechanism [46]. This approach is taken from the author's findings that the torque of the toe during the stance phase is almost proportional to the torque of the ankle for the latter 40% of the stance phase [46]. It is also derived that because the velocities of each joint are of similar magnitude in opposing directions, the toe dissipates power, and the ankle generates power [46]. Due to this, it is assumed that this information would allow both the toe and the ankle to be powered by a single actuator, which is beneficial in reducing both mechanical complexity and weight [46].

The ankle joint is an underactuated system, which is implemented in a five-bar mechanism providing 2 degrees of freedom in the sagittal plane [45]. The prosthesis is driven by a prismatic joint, which is actuated by a linear series-elastic actuator that provides motion to the ankle and the toe simultaneously [46]. This is realised by a pin joint driven by three 45° bevel gears, the first of which is mounted to a helical gear which is driven by another helical gear connected to the motor [46]. The second bevel gear is mounted freely to the case of the shank, which transmits the rotation of the motor 90° and allows for ankle actuation [46]. The final bevel gear is mounted to the secondary gear and connects to a ball screw, which runs down the length of the foot, which allows a linear spring in the toe to be compressed and decompressed [46]. The other end of the linear spring is connected to the penultimate linkage, which connects to the toe via a sagittal plane pin joint. The toe is then mounted to the body of the foot via another pin joint [46]. Compressing or decompressing the spring applies a force on this linkage chain, which in turn drives the toe joint [46]. This configuration means that the torque of the ankle joint is greater than the torque of the toe at any given force generated by the motor [46].

Work has been done in developing 2-DOF ankle-foot prosthetics. Ficanha et al take an approach to a 2 DOF, by designing a system predominantly in the interest of implementing a new control scheme [8]. The prosthesis itself consists of an Ottobock Axtion foot, attached to the shank via a universal joint padded with an elastomer providing passive resistance similar to Xiu et al, as shown in Fig. 10[50]. The ends of a nylon rope are attached at the back of the foot on both sides of the longitudinal axis; both ends then wind onto rotating drums mounted on either side of the shank, then the middle of the rope is attached to a pulley mounted to the front of the foot [8]. This means that inversion and eversion can be applied with opposing motor rotations, and dorsiflexion and plantarflexion with the same motor rotation [8].

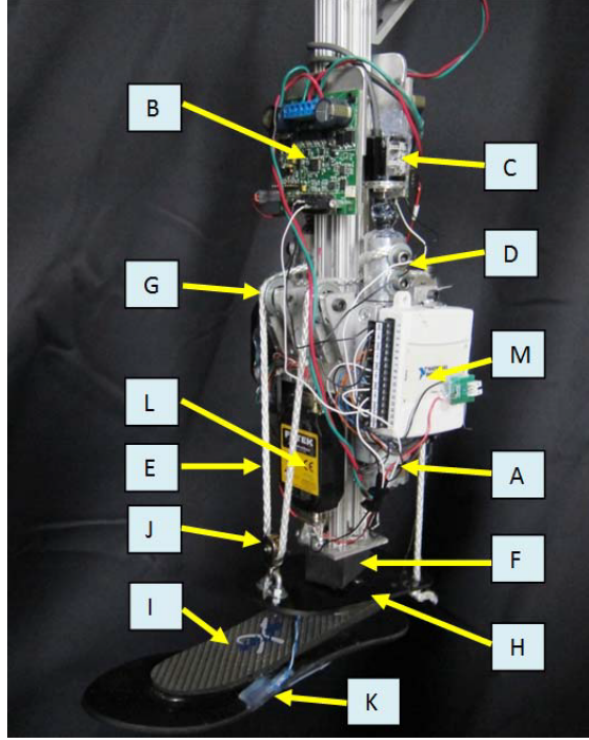


Figure 10: The Ficanha et al 2 DOF ankle implementation. Taken from [8].

Strain gauges on the foot, in a Wheatstone bridge configuration, provide torque feedback [8]. The 4 strain gauges are mounted in pairs in front and behind the centre of rotation of the sagittal plane [8]. A voltage drop at the output of the Wheatstone bridge indicates a force at the heel, while a voltage increase indicates a force at the toe [8]. This can be used to infer initial contact and toe-off [8]. At foot flat b, both effects cancel, so the output of the Wheatstone bridge can be considered to be representative of the net ankle torque [8]. For inversion and eversion, another Wheatstone bridge is employed, of which the pairs are mounted on the top of the foot, and the others to an inert carbon fibre plate [8]. Different strains in the coronal plane will cause the voltage of the Wheatstone bridge to increase or decrease, which can be correlated to the torque in eversion and inversion during ground contact [8].

The implemented control system is inspired by the observation that at initial contact, the ankle accepts the environment and is manipulated by external force, while at push-off,

the environment now accepts the ankle as it generates its own desired motion [8]. This means that, through the initial contact, the ankle can be considered to be in admittance and at push-off in impedance, which would suggest the prosthesis can be controlled by an admittance controller at initial contact and an impedance controller at push-off [8]. This is in contrast to Xiu et al, who tuned the rubber block to provide static ankle stiffness [50]. Here, Ficanha uses a uniform block and can dynamically control the stiffness. This control scheme was implemented in a finite-state machine using pre-recorded angle data in the frontal and coronal planes [8]. Starting in mid-swing, the controller starts the system in impedance control, where it will remain contingent on index I , the index of the pre-recorded data, reaching initial contact or if initial contact is detected itself [8]. Moving to a secondary initial contact check, the system will start a loop in the case initial contact is later than expected to account for the variability of gait, after which it shifts into admittance control [8]. It will remain in admittance until I reaches the foot flat point, allowing impedance control for push-off [8]. Finally, the state machine cycles back to the start when I indicates the swing phase [8]. An experiment was designed to test the prosthesis capability under turning; a passive prosthesis is lacking in this functionality, as it requires both motion planes, resulting in increased forces in both planes [8]. Using a circular treadmill, Ficanha et al demonstrate that the device significantly reduces the torque in both planes compared to a passive prosthesis, which would result in better and more stable mobility [8].

Li et al. demonstrate a 1 DOF biomimetic foot-ankle prosthesis, which is an example of a non-pin-jointed ankle, shown in Fig. 11. Li et al. recognise the clear kinematic and kinetic discrepancy between a human synovial joint and a pin joint, stating that a pin joint or rigid link joint compensates for its simplicity with good material properties [25]. However, this compensation is limited due to the fact that it is mechanically unable to replicate the translational component of the rolling joint movement, meaning they simply cannot match the trajectory of the human ankle [25]. Li et al proposed a design

for a joint representing the tibiotalar joint, inspired by the bouncing joint structure of insects. The trajectory of the tibiotalar joint was taken from data points of a previous study, characterising it as a "J" curve, which was used to calculate the geometry of the contact surfaces [25].

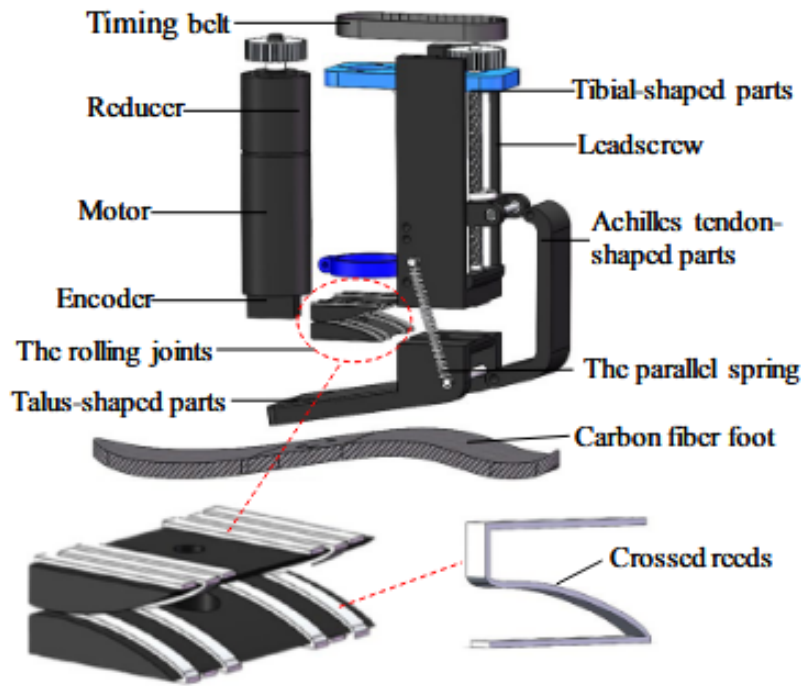


Figure 11: The Li et al design showing the rolling contact joint implementation. Taken from [25].

This is then paired with sprung crossed creeds to support and restrict the motion of the joint to the sagittal plane [25]. Further building upon the biomimetic approach, the actuation is provided by a linkage, similar to the Achilles tendon [25]. The artificial tendon is driven by a vertically mounted motor in the shank via a lead screw [25]. The device uses a carbon fibre foot for energy storage, providing windlass-like action, much similar to various foot-ankle prostheses [25]. The toe trajectory test of the device matched the toe trajectory of the human foot [25]. The demand for dorsiflexion and plantarflexion is also met, and the energy storage of the foot is 88.96%, which could theoretically reduce

metabolic costs [25].

The SPARKy project by Bellman et al aims to reproduce the human gait cycle efficiently using regenerative kinetics [3]. Both SPARKy 1 and 2 are powered 1 DOF foot-ankle prostheses, shown in Fig. 12, incorporating a vertically mounted motor at the rear that drives a carriage, springs mounted on either side[3]. These springs are then connected to a lever, and then the foot platter, which is free to rotate in the sagittal plane in a pin joint [3]. This allows the motor to drive plantarflexion and dorsiflexion, with the springs providing energy storage [3]. SPARKy 2 has a similar configuration to SPARKy 1, but aims to improve the design using a short lever arm and a higher output motor, making it lighter and smaller than SPARKy 1 [3]. SPARKy 3 applies the SPARKy 1 & 2 methodology to a 2 DOF version of the prosthesis [3]. The design used two of the same motors from SPARKy 2, as the new degree of freedom needs to be actuated, which are also mounted vertically at either side of the shank [3].

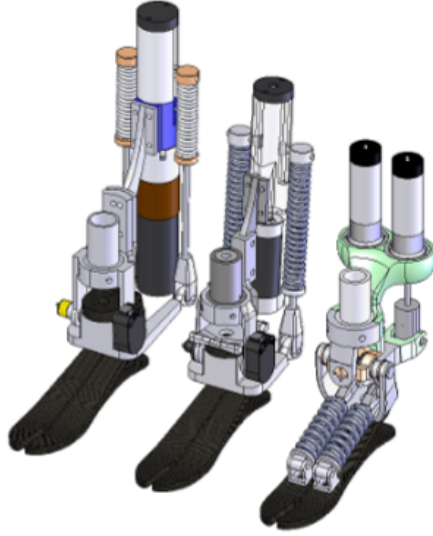


Figure 12: All three generations of the SPARKy project from 1-3, starting from the left. Taken from [3].

The motors are mounted to the shank via ball joints, which allow them to rotate under actuation in the coronal plane, in which they are restricted [3]. The sprung linkages are gone, replaced with the motors directly connected to two L-shaped arms via a lead screw

[3]. Both of these arms connect to a lever in the middle, which is directly mounted to the back of the foot, via a pin which forms the sagittal rotation axis. The other ends of the L-arms connect to two springs, which are mounted to the forefoot, providing the energy storage that the vertical spring provided in SPARKy 1 & 2 [3]. A universal joint is formed by a rotating part on the sagittal plane pin in between the L-arms, which has a second rotating axis perpendicular to the sagittal plane, forming the coronal plane axis [3]. While the sagittal axis is unrestricted, the coronal axis is restricted at ± 20 degrees by hard stops on the rotating pin [3].

Jang et al. explore a 2 DOF foot-ankle prosthesis, shown in Fig. 13, which also uses two rear-mounted linkages similar to SPARKy 3 [20]. The design here uses two harmonic drive actuators horizontally mounted to the shank, rotating in the sagittal plane [20]. They both have a lever arm which connects to two parallel links via a pin joint. On the other end of the link, they connect to the heel of the foot on either side using a universal joint [20]. The foot is then connected to the ankle via a universal joint, which has a wedge-shaped protrusion in the coronal plane [20]. The wedge contacts the joint to limit the rotation in the coronal plane to ± 20 degrees, the same limits as SPARKy 3 [20]. The range of motion for the sagittal plane is ± 30 degrees [20]. In parallel with and between the two actuated links, there is an inline spring mounted via a rotational axis in the sagittal plane shank and connected to the middle rear of the foot; this link isn't actuated [20]. This spring only compresses in a dorsiflexion movement, when a stopper engages with the spring, enabling compression and storing energy [20].

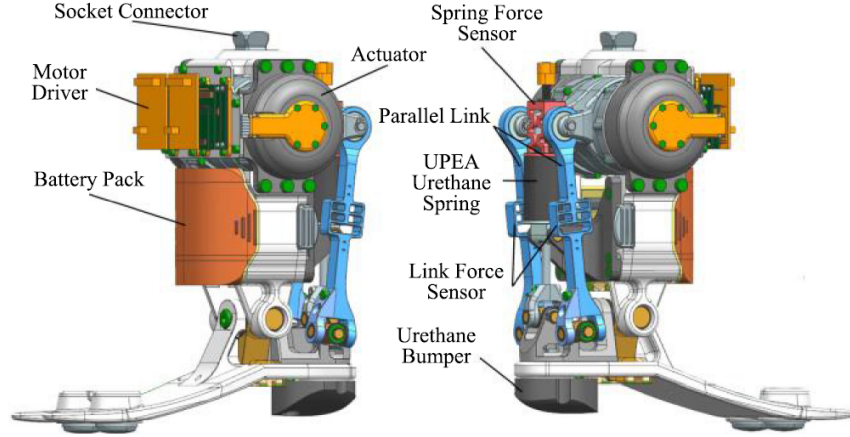


Figure 13: The Jang et al ankle implementation showing the sensorised parallel links. Taken from [20].

The novel idea for this design is the inclusion of 3 capacitive sensors in the rigid links of the device, to measure the ground reaction forces of the foot [20]. There are two sensors in the design; the first one is on the central spring link, which is mounted as a buffer piece between the spring and the spring stopper [20]. The sensor is rated to 900 N and can measure the force with 5.44% error [20]. The other two sensors are integrated into the actuated parallel links, where the capacitive plates are mounted inside a hollow protrusion in the middle of the links, rated to 800 N with a 2.79% error [20]. The idea here is to infer the forces of the foot from the links [20]. There is an emphasis observed on displacing the weight of a prosthetic foot up into the shank, as it is both more efficient to drive a lighter foot and more like the anatomy [20]. Ficanha et al. have the force sensors on the foot, increasing its weight. Here, Jang et al offload the sensors to the actuation links. It was confirmed that inferring the forces from the links is an accurate alternative. This displays the ability to use the successful control concepts from Ficanha et al with the mechanical design of SPARKy 3 in a novel way, producing a more advanced prosthesis.

2.7 Robotic Sensing and Artificial Ligaments

Proprioception is the way the native body is aware of its position without visual feedback. Traditionally, in current prosthetics, this is done with sensors such as encoders to estimate joint angle and, therefore, infer position. To take a biomimetic stance on design, it is important to understand different sensing methods that could be applied in the context of biomimetics to provide a similar result.

2.7.1 Artificial Ligaments

The LARS ligament is a medical device used in vivo to replace ligaments after injury [28]. Its primary application is to act as a scaffold for ligament regeneration, whilst maintaining the mechanical properties of the ligament while regeneration takes place [28]. First made available in 1992, the LARS is made out of polyethylene terephthalate (PET) fibres and has been commonly adopted for knee reconstruction [28, 14]. The LARS has also shown a lower failure rate compared to its predecessors [28]. However, a study by Wang et al proposes the use of polyetheretherketone (PEEK) as an alternative material [48]. PEEK exhibits mechanical properties that would be excellent for a ligament replacement device, including resistance to wear and fatigue [48]. In the context of the study, the interest in PEEK comes from its excellent osteogenic activity, which PET does not have [48]. Purely from a mechanical perspective, PET, having been more widely studied, is an acceptable choice ex vivo when compared with PEEK [48].

2.7.2 Stretch Sensing for Artificial Ligaments

Traditional strain sensors are mounted on a rigid surface; this technology would not function correctly on a flexible part. Flexible systems require a more specific configuration to work correctly, which is most commonly found in two technologies: resistive and capacitive [41]. Capacitive sensing operates with the capacitance between two electrodes varying with the deformation of the flexible part, due to the distance of the electrodes changing under deformation [41]. Resistive technology operates by using the piezoresis-

tive effect, in which the deformation of the electrodes under strain causes a change in the electrode resistance [41]. Resistive sensors tend to have a high sensitivity, quantified as gauge factor, but they show a large hysteresis and are non-linear [41]. Sensors with capacitive technology tend to have lower sensitivity but are linear and have small hysteresis [41].

While resistive and capacitive are two major modes of operation for strain sensors, there are other strain sensors using different technologies. Optical waveguides have been implemented in the soft robotic domain in an effort of trying to achieve reliable and low-power soft sensors for use in human interaction [19]. These devices operate according to the principles of traditional hard fibre optic cables, usually made of glass or plastic, but are made of soft materials instead [19]. The soft materials allow the deformation of the waveguide structure, an optically transparent core with a cladding, to be exaggerated beyond the range that hard materials could reach. This allows deformation of the material to be characterised, but the resulting optical power loss or gain as the length and core cross-section decrease or increase [19]. Jamil et al. identify other strain sensor architectures that operate under the optical mode of measurement, but differ in operation from measuring a linear deformation directly with powerless [19]. To et al achieve a soft optical waveguide using a homogeneous rubber substrate, using a gold film as a barrier to allow total internal reflection within the core [44]. The mode of measurement is still the optical power loss; however, in this case, as the device is deformed, micro-cracks form in the gold film, allowing light to leak out of the waveguide and cause an intensity loss [44].

Yun et al. employ an optical strain sensor architecture similar to that of traditional hard resistive strain sensors by embedding a plastic optical fibre in a PMDS matrix in a serpentine shape [51, 19]. The optical fibre is thermally set into its shape, applying several corner radii to the path of the fibre at rest [51]. Total internal reflection is dependent on the incident angle, which decreases with an acute bend, scattering the light. As the fibre is stretched, the radius of the bend increases, which makes the incident angle

larger, decreasing the power loss [51]. Harnett et al follow a standard linear waveguide strain sensor design, simply using a straight waveguide with a urethane core and silicone cladding [11]. They use a 600 nm wavelength with an optically transparent core and cladding. This device was stitched into a fabric material, which was used to measure the size of the hamstring muscle, which is deformed around a curved surface and functions with a clear signal [11]. Finally, the resulting design based on the literature search conducted by Jamil et al follows a similar principle to Harnett et al, with the design using a urethane core and a silicone cladding [19]. The novelty in this design, however, was the use of infrared light and the inclusion of a pigmented cladding to help with noise reduction.

This ultimately improved the signal-to-noise ratio of the device, and therefore validated the use of IR light in the waveguide, which reduces the power consumption of the device [19]. The soft waveguide was found to be multimodal, capable of sensing linear stretch, bending, and compression reliably and with high repeatability [19]. Compared to the other waveguides reviewed, the method to reduce the power loss appears more similar to traditional methods and appears easier to manufacture. This architecture could then be a suitable method to act as a soft and proprioceptive artificial ligament.

2.8 Previous Works

The work to date surrounding this project was started by Russell et al with the design of a sagittal plane restricted mechanical bicondylar knee joint, shown in Fig. 14 [40, 39]. The design of the knee is similar to the Li et al process, but began by taking a 3D scan of the femur and isolating the articulating surface of the femoral condyles, which was then simplified [39]. A computer model was then derived to calculate the matching surface of the tibial condyle, ensuring constant rolling contact during motion [39]. The joint is constrained using an analogue of the anterior cruciate ligament and the posterior cruciate ligament [40]. The ligament design uses two mounting blocks, each with a pivot for the femur and tibia, with a rod attached to the tibial block, which then freely passes through

the femoral block [38]. On the other side of the block, a spring is placed on the rod, and a nut caps it off, providing the elasticity of the ligament, which was chosen to be a 126 N/mm spring as it was identified to match the anatomy [38]. A test rig was built to experimentally validate the design. This had the knee built using stainless steel for the tibial condyles and plastic for the femoral condyles to reduce friction. To measure the stretch of the ligaments, LVDTs were employed, with the core mounted on the tibial pivot block and the housing on the femoral [38, 39]. The knee was driven on a squatting simulator, with two actuators acting as the quadriceps and hamstrings [39]. Results from testing on the rig showed that the ligament stretch, rotation and motion of the knee matched that of the predicted model, and by extension the human knee [39].

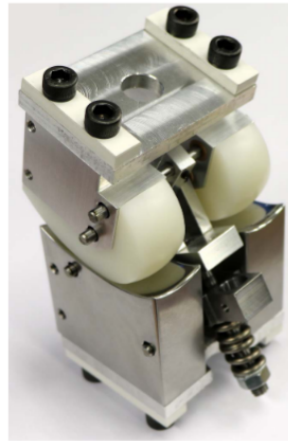


Figure 14: The Russell Knee joint. Taken from [37].

Building upon this concept, MEng student Thomas McMullen undertook his dissertation expanding Russell's initial ideas, shown in Fig. 15. McMullen aimed to assess the detection of various motions with a bent knee rig using the Russell knee joint [43]. The aims were to understand the effects of walking speed variation and intent detection from quiet standing to walking [43]. The evidence showed that there were no markers of intent 1 second before the first toe-off initiating gait in joint angle and the displacement of the ACL and PCL [43]. However, it is stated that the quiescent standing state is distinct from other events, although there is significant noise in the signal due to small movements. [43]. Therefore, while detection of intent before the event is unachievable, the

clear detection of a state change is possible [43]. For the speeds 0.53 ms^{-1} , 0.68 ms^{-1} , 0.86 ms^{-1} , 1.05 ms^{-1} , no statistically significant variation in speed was shown.



Figure 15: The bent knee rig implemented by both McMullan and Leake. Taken from [43].

The author undertook their own BEng dissertation, expanding on McMullen's work, in the event detection of speed using the Russell knee joint [17]. This project aimed to focus specifically on identifying a variation in speed in the ACL and PCL signals, with the addition of surface electromyography (sEMG) [17]. A more robust bent knee rig was designed and fabricated for consistent and reliable testing, shown in Fig. 16. Although gait cycles were previously manually isolated during processing, a sensor on the foot heel was installed to isolate cycles [17]. Although sEMG and ligament displacement testing were not performed simultaneously, it was still shown that there was a clear proportional variation in speed in ligament displacement and sEMG signals during gait, with correlation between speed-dependent events and the data sources [17].

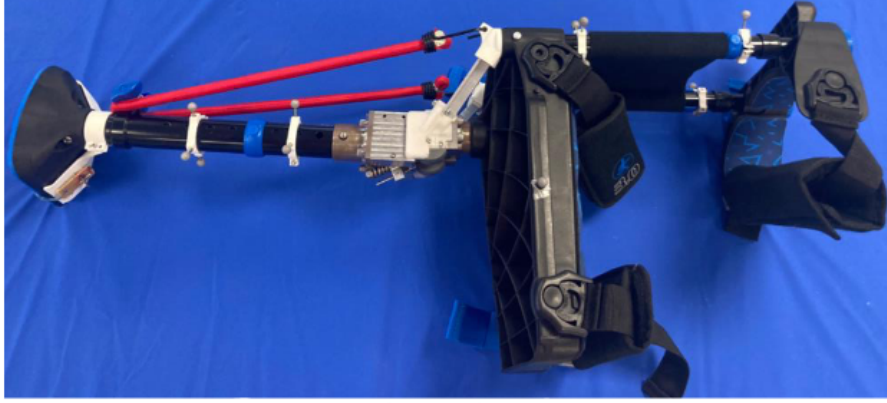


Figure 16: The Bent Knee rig implemented by Clark. Taken from [17].

2.9 Biological Comparison

2.9.1 Ankle Motion

The primary role of any prosthesis is to provide the function of the missing portion of the body; the ultimate goal should be to have the device function as the missing body part would. This requires mimicking the anatomy as closely as possible, the principle of biomimetics. It is an intuitive concept that varying a variable of a system will ultimately affect the system as a whole. This is the motivation for the previous works on this project: simplifying the complex kinematic characteristics of a knee joint to a pin joint will ultimately deviate functionally from the human knee. The bent knee rigs of the previous projects had a rigid foot ankle non-articulating ankle design [17, 37, 7, 43]. there could be a benefit to implementing an ankle design into the system, using the bio-mimicking design sentiment of the Russell knee joint [17].

There are some interesting merits to the implementation of the prosthetics reviewed in the previous sections, specifically in the scope of biomimetics. The design by Li et al shows that there is a clear benefit to a bio-inspired design implementing a rolling contact joint in the ankle [25]. As the tibiotalar joint is a convex and concave interface, rather than the two convex facets in this design, this might reduce the stability of the joint, as

the tendon and mortise structure of the tibiotalar joint is crucial to its stability [25]. Li et al however, were able to match the toe trajectory of the human foot with the bouncing joint design due to the added translational component of its movement, like the human joint, unachievable by a pin joint [25]. This design is restricted to the sagittal plane and therefore only has 1 DOF, Ficanha et al has shown that implementing 2 DOF is critical to balance and stability, especially when turning as a 1 DOF device is not able to permit motion in the axis of turning inducing more stress and force on the user and prosthesis, of which the Ficanha et al design has shown to reduce [25, 8].

Most interesting is the observation made by Ficanha et al in regard to the admittance and impedance states in the ankle, which ultimately inspired them to implement this into the control system [8]. This concept showed that, in conjunction with the 2 DOF, this control scheme can significantly reduce the torque on the system in motions in the coronal plane, which ultimately increases mobility and balance [8]. As we have seen previously, the superior articulating facet of the talus has a taper posteriorly [5]. Therefore, in dorsiflexion, the joint is very stable, with little motion in the other planes permitted [5]. However, in plantarflexion, where the taper is most active in the joint, the joint becomes more compliant due to the reduction in surface area at the facet interface [5]. This behaviour appears to correlate with the observation made by Ficanha et al, as at initial contact, the ankle is in a marginal plantarflexed, effectively neutral, where the tibiotalar joint is stable, and the ankle is in compliance [8, 5]. On the other hand, at push off, the foot is in plantarflexion; therefore, it is compliant due to the taper of the talus and is in impedance [8, 5]. Therefore, the control system designed by Ficanha et al is directly mimicking the kinematic behaviour of the tibiotalar joint in its control system. The joint is rigid when it makes impact with the ground, allowing the ground to manipulate the joint, and is compliant when the foot being controlled manipulates the environment [5, 8]. Fincha is achieving this behaviour with software, implemented in the design, which uses a universal joint, which has the same disadvantage of simplified kinematics [8]. It has

been shown by Xiu et al that soft passive ankle kinematics are possible and do have a positive impact on gait, capable of approaching a normal gait cycle. Unfortunately, its passive design gives it the issues of unpowered prosthetics that powered designs aim to solve [50].

2.9.2 Analogue of Ligaments

The capacitive strain sensors employed by Jang et al were an attempt to measure ground force reaction in line with the actuation system, by measuring the compression force on the linkage due to the ground reaction force [20]. The Golgi tendon organs are contraction-sensitive mechanoreceptors, found in the muscle-tendon interface in a collagen capsule and are innervated by fast-conducting afferent fibres [18]. They work in conjunction with the muscle spindles, located in the core of the muscle body, to provide constant monitoring of the muscle state [18]. Where the spindles provide information about length, the tendon organs provide information on tension [18]. This provides another similarity: if the linkages of the Jang et al design are considered analogous to the tendons of the ankle, then the capacitive sensors can be thought of as representative of the tendon organ [20, 18]. The soft tissues of the body are heavily innervated, not only the tendons, but also the ligaments of the human body [12, 34].

Both the lateral and medial ligaments of the ankle joint are innervated, specifically the lateral ligaments being noted to be highly innervated with mechanoreceptors alongside the joint capsule of the tibiotalar and subtalar joints [12, 34]. This includes Golgi tendon organs, but it would seem that the mechanoreceptors with the highest distribution are Ruffini, which work in a similar principle to Golgi organs [34]. The ATFL had significantly more innervation than the medial ligaments, which further shows the lateral ligaments provide more sensory information than the medial ligaments [34]. In principle,

the innervation of the ligaments works similarly to that of the tendons; theoretically, the concept of measuring the strain on analogue tendons could be applied to analogue ligaments. This concept runs in parallel with the design of the Russell knee joint, which primarily aimed to also gather feedback from the stretch of the ligaments. Renstrom et al studied the strain profiles of the lateral ligaments of 5 cadaveric ankles, for the ATFL and the CFL, shown in Fig. 17 and Fig. 18. Testing was done primarily to see how the strain of the ATFL and CFL differs in the motion of dorsi-plantarflexion across differing foot positions [35]. A general conclusion was drawn that the stretch of the ATFL and the CFL are mutually exclusive in nature, meaning that the gradient of the two profiles will be counter-opposing [35]. There is a crossover point at 0° or the transition between dorsi-plantarflexion [35]. This then gives a distinct pattern for both sides of the motion plane that could be used to identify if the joint is in either dorsiflexion or plantarflexion. Using artificial ligaments capable of measuring their own stretch could then provide proprioception in a way similar to both the Russell knee joint and the anatomy of the mechanoreceptors in the lateral ligaments [40, 34].

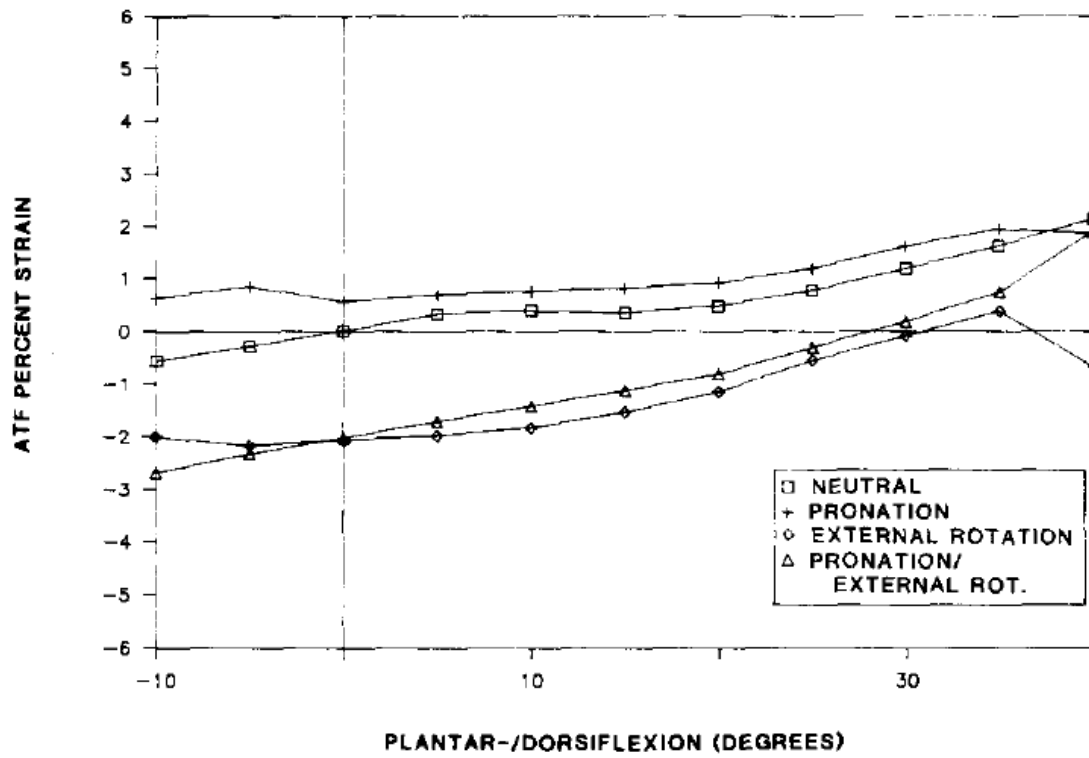


Figure 17: Taken from [35]

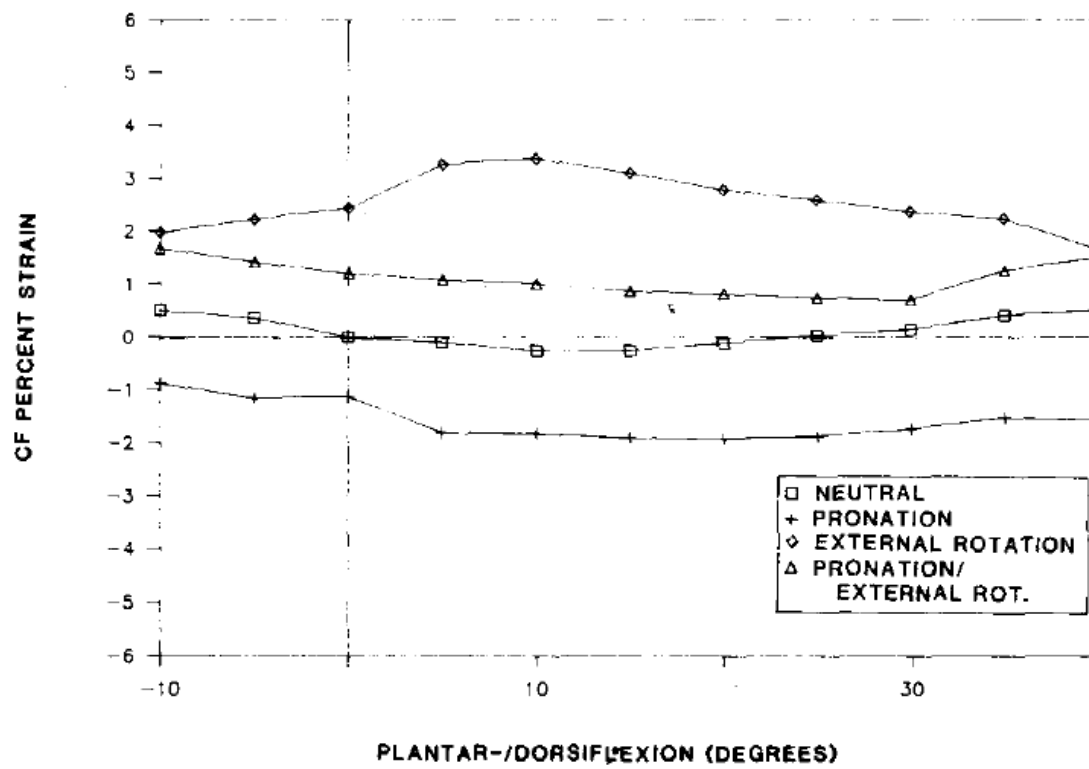


Figure 18: Taken from [35]

2.9.3 Summary

In summary, the local anatomy of the human ankle has been explored, identifying the key joints of the ankle complex and the planar motion they provide. The tibiotalar joint is the primary contributor towards sagittal plane movements, and the subtalar joint is the major axis for providing motion in the coronal plane. Ligaments restricting motion in these joints have been identified as the anterior talofibular, calcaneofibular, tibiocalcaneal, superficial tibiotalar, and the interosseous talocalcaneal ligaments. It has been observed that the strain in lateral ligaments could provide proprioceptive feedback to the motion of the ankle joint; therefore, a review of deformable strain sensors was conducted. Optical waveguide strain sensors were identified as the most effective and reliable method of measurement with a deformable strain sensor.

3 Aim and Objectives

This project aims to produce a biomimetic rolling contact ankle joint complex from 2 sub joints. This joint will be constrained with deformable waveguide-based ligaments, which can provide feedback about its own deformation. The aim is to provide human-like motion in the joint as opposed to the traditional pin joint. The two discrete joints provide a further range of motion to increase balance and stability. The deformable waveguide-based ligaments could be used collectively to estimate the planar position of the sole of the foot. Based on the review of the current literature, the following aims and objectives have been laid out to design, test, and validate the bio-inspired ankle joint capable of proprioception:

1. To simplify the geometry of the ankle joint to design and fabricate an idealised joint with two sub-joints enabling motion in the sagittal and coronal planes.
2. To design and develop deformable waveguides that operate to the 6% strain required for use as an analogue proprioceptive ligament.
3. To test the ankle joint in a constrained environment to verify that it can achieve 15° dorsiflexion, 45° plantarflexion, 25° inversion, and 15° eversion.
4. To test the ligaments in the system to verify if unique patterning in the linear displacement can be identified, to find the planar position of the joint.

4 Materials and Methods

4.1 Ankle Design

4.1.1 CT Data

In order to produce an ankle design that closely matches human geometry, originating from in-vivo human bone geometry is a fundamental starting point. For this reason, a high-quality Computed Tomography (CT) scan was used. The initial criteria for selecting a scan were a healthy adult. The virtual skeleton database (VSD) is a large dataset of whole-body postmortem CT scans. It consists of 30 different subject data sets, including 20 full scans and 10 inconsistent or incomplete [23, 9]. Patient 17 appears to be an ideal data set to be used as an anatomical reference for the topology of the ankle. The available data set for Patient 17 has clear, high-quality images, with a voxel size of 1 mm and good contrast between soft and hard tissues, shown in Fig. 19. Patient 17 is also reported to be 19 years old with average height and weight. [23].

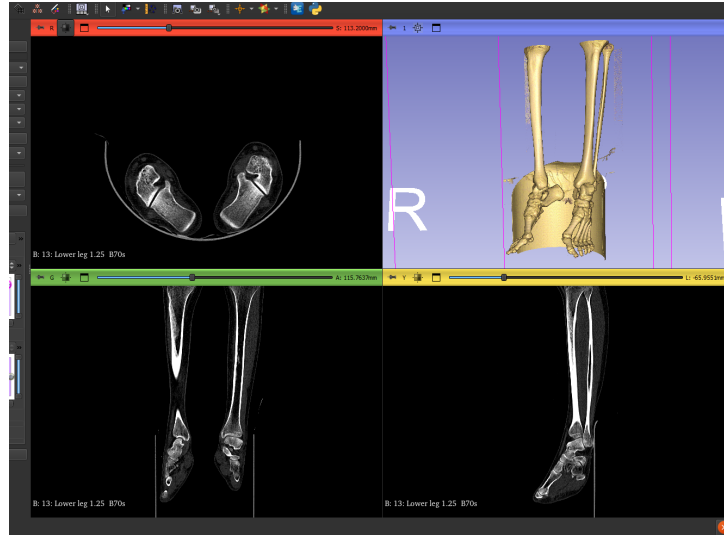


Figure 19: The 3 plane DICOM file of Patient 17 viewed from 3D Slicer showing the local bone structures.

4.1.2 Segmentation

In order to best take advantage of the dataset, having a three-dimensional geometry would prove useful in terms of being able to readily measure and edit the topology of the bones. This would mean post-processing the CT scans to extract the bones as 3D CAD files, which then could be exported into a standard CAD software. The most common method of achieving this is known as segmentation, where individual pixels in a scan are assigned 3D pixels, or voxels, to build a 3D model of an object. 3D Slicer is an open-source medical imaging software that contains a toolbox that allows for the segmentation of CT scans. The virtual skeleton database was used in Fischer et al, aiming to isolate the geometry of the full lower skeleton, also using 3D Slicer [9]. The workflow of their isolation of the data set was reported, as using a threshold of 200 Hounsfield units for initial segmentation of the bones [9]. Once imported into 3D Slicer, a threshold of 200 Hounsfield units was applied to the scan using the software threshold tool, which isolated a majority of the bone structures from the scan [9]. The threshold left significant artefacts in the 3D model, resulting in the need for manual clean-up.

The whole skeleton, omitting the right lower tibia and foot, was manually removed. Then the island tool in 3D slicer was used to remove small noise-like artefacts, likely from high-density areas of the soft tissues. This resulted in a rough model of the lower shank and the bones of the foot, which, due to the imperfect nature of a threshold, had varying degrees of remaining surfacing imperfections, joint fusion and cavities where the trabecula was removed during the threshold. A combination of the tibiotalar joint and the subtalar joint would be capable of providing a minimum of 2 planes of motion [5]. Primarily, in this case, sagittal and coronal plane motion, providing dorsi/plantarflexion and ev/inversion [5]. The forefoot in the 3D slicer was removed, with the shank and hindfoot remaining. Sections of the cuboid and navicular bones remained due to joint fusion between the talus and calcaneus. Joint fusion was also present in the subtalar and tibiotalar facets, which were manually removed by painting out all of the connecting

spurs, leaving the fibular, tibia, talus, and calcaneus bones isolated, with a rough surface.

The voids and imperfections of the mould remained, including the rough geometry of the facets. The island function could be used inversely to generate a new structure from a negative space. As this was considered an *island*, it would need to have a clear boundary in the model. Therefore, all surface voids were manually filled using the painting tool, which was inserted depressed from the surface of the model so as not to interfere with the surface topology at this stage. The island tool was then used to generate new 3D projections of the voids inside the bones, which were then merged with their corresponding bones to make solid bodies. This left the surface imperfections of the bones, which were handled by manually painting, erasing and smoothing any obvious depressions and spurs present. This was done by referencing the original scans to check that the manually edited geometry matched that presented in the scans. Finally, the default gross body smoothing filter was applied to all the bones, as the model before smoothing is only a voxel map, a collection of cubes; therefore, the smoothing is necessary to match the smooth geometry of the bone, shown in Fig. 20.



Figure 20: The segmented, cleaned and smoothed mesh object of the fibula, tibia, talus and calcaneus exported as a .stl from 3D slicer.

4.1.3 Initial Cad Setup

The model containing all the segmented bones was imported into a conventional CAD software as a .stl file. Imported as a mesh, the surfaces were not watertight; therefore, they were stitched, and the mesh was converted into a standard body, separating the bones into 4 discrete bodies. These bodies did not only include the geometries of the bones, but an anatomical position too, a conscious choice to aid design. The analogue for the shank was chosen to be a length of 40 x 40 mm extrusion, which means the tibia and fibula had to be considered fused. Therefore, the distal tibia and fibula isolated in this

project should also be considered fused. A portion of the non-articulating bone, where the tibiofibular interosseous membrane would be, was connected to preserve the positions of the bones. At this stage, all three bones were fabricated in 3D printed PLA to gain a tactile understanding of the motion facets, range of motion, and contact points.



Figure 21: The smoothed bone structured as 3D printed from the CAD files. Shown here from left to right are the now fused shank (the tibia and fibula), talus, and calcaneus bones.

4.1.4 Facet Isolation

Simplifying the ankle joint meant isolating the key motion facets from the bulk of the bone, allowing these geometries to be idealised, separate from the bulk of the bone. The anatomical positions of the facets were then preserved by means of a skeletal-type scaffold. The facets of interest in the shank are those that allow the motion of the tibiotalar joint. Preserving both lateral malleoli is critical to the stability and function of this joint, as well as the top articulating surface, constraining the superior surface of the talus. The fibula and tibia were then cut at the most distal point of the malleoli, and the top

articulating facet was cut at the point where the tibia and fibula fully form the dome with extra margin for rigidity, shown in Fig. 22. These surfaces were then kept, and the rest of the bone disregarded. A mount was then designed, holding both the tibia and fibula motion facets in place, and also serving as the attachment point to the 40 x 40 mm extrusion.

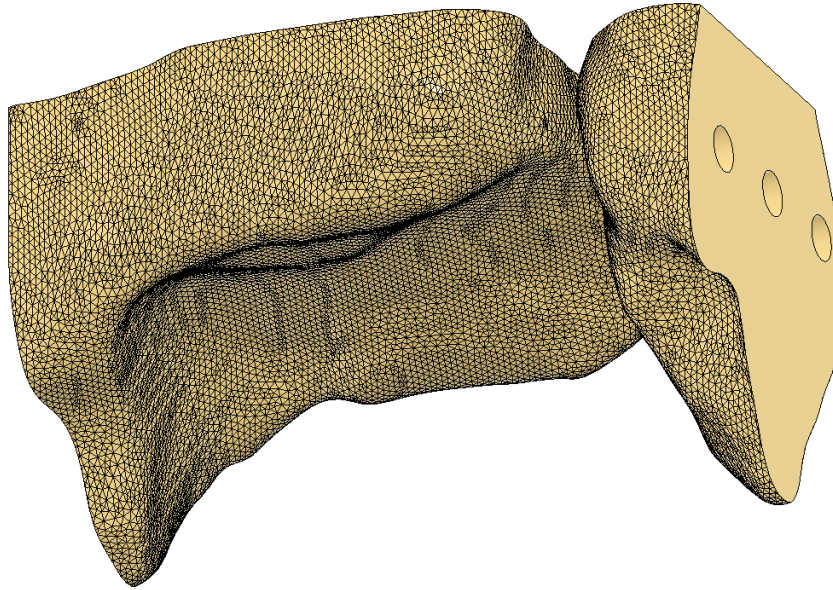


Figure 22: The isolated distal tibia and fibula motion facets.

The calcaneus is the structural heel of the foot. Therefore, only its anterior and superior sides have motion facets. For this reason, only the superior anterior section of the calcaneus needs to remain, as the distal anterior facet meshes with the cuboid bone, which is out of the scope of this project. The superior anterior section is primarily the calcaneal motion facets of the subtalar joints, and was kept, removing the rest of the structure, shown in Fig. 23. The remaining structure of the calcaneus was replaced by a mounting structure acting as the heel. Sandwiched between the calcaneus and the tibiofibular complex, the talus is responsible for the tibiotalar joint superiorly and the subtalar joint inferiorly [5]. This makes the facet isolation of the talus a more complex structure than the previous two bones, as it has to have a centre scaffold holding the superior and inferior facets in their anatomical positions.

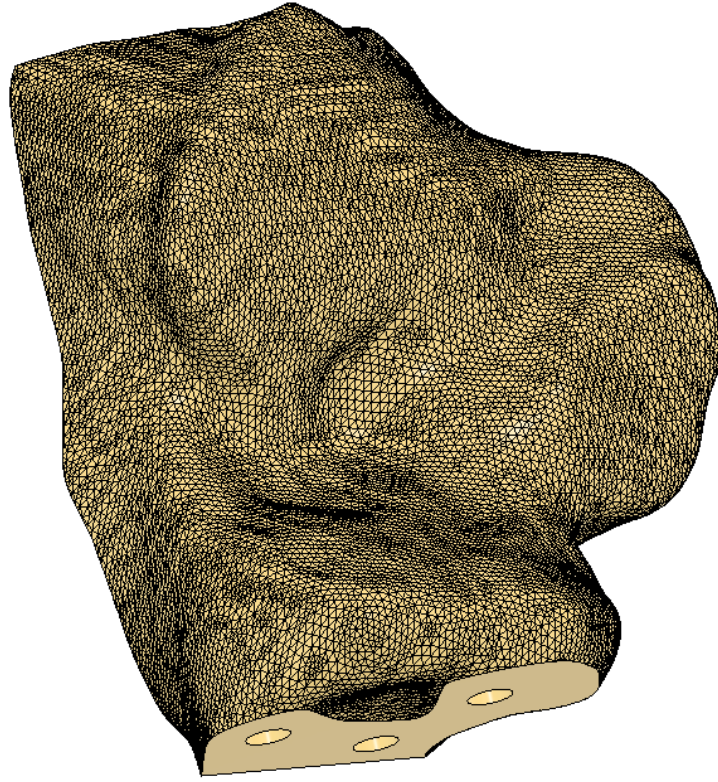


Figure 23: The isolated anterior superior calcaneus facet structure.

Superiorly, the trochlea of the talus is the main articulating surface in the tibiotalar joint. Also to note is the bony depression at the anterior of the trochlea [5]. What is interesting in this case, is that the depression forms a notch that the anterior edge of the tibial dome interferes with, acting as a hard stop. To preserve the lateral and medial rigidity, a portion of the lateral and medial sides of the talar trochlea was kept to allow the surfaces to restrict out-of-plane motion. The talus is then cut to keep the facets to this specification. For the inferior articulating surfaces, there are two key facets to the subtalar motion. The posterior facet and the anterior gliding surface, both of which have discrete borders and are isolated, and the rest of the bone is removed. The scaffold to hold the facets was then designed, gripping the superior facet, then extruding down with a central strut, which holds the plate surfaces that grip the inferior facets.

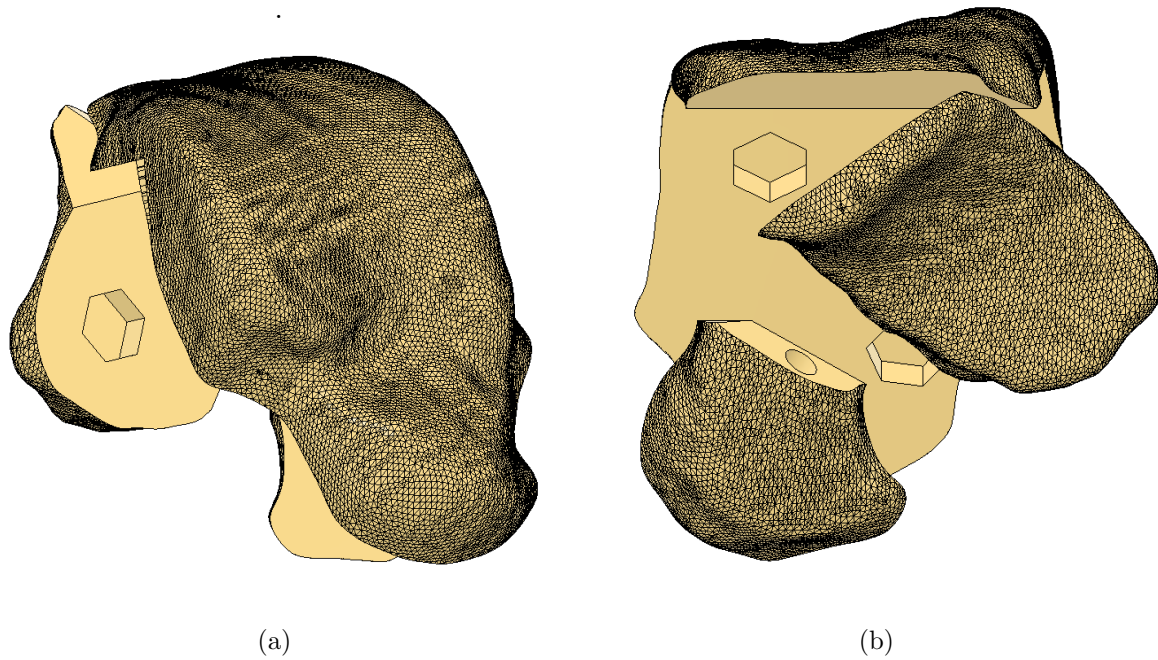


Figure 24: The isolated joint facets of the talus, with the superior facets (a) and the inferior facets (b).

4.1.5 Facet Simplification

The geometry of the bones was segmented directly from a CT scan, and it does not come without drawbacks. Soft tissues are difficult to differentiate in a CT scan, leading to the bone segmentation being without the cartilage structures. These are the true contact surfaces of the joint, and therefore, the bone is only representative of the geometry. To counteract this, the joint surfaces can be simplified to a simpler and more easily manufactured form. Another reason for this is to have control over the planar locking of the joints. Both the tibiotalar and the subtalar joints show signs of having multiplanar motion; planar locking of the joints could improve control and increase stability.

For the tibiotalar joint, the geometry of the trochlea of the talus was simplified. The geometry in the sagittal plane is the most important to match the motion of the joint. Secondly, the groove shape in the trochlear in the frontal plane provides some degree of registration. The surface was bisected in the middle of the facet, and a spline was

then matched to the curve in the sagittal plane. The trochlea is then also bisected in the frontal plane, and a second spline is taken to match the curve of the groove. Using the sagittal plane curve as a guide rail, the frontal plane geometry is extruded along the guide path to form the simplified facet. The anterior protrusion of the talus was simplified by matching a radius to the protrusion. The shank was simplified by continuing the fusion of the fibula and tibia. The shank mount has already removed any motion between the articulating facets; therefore, it was advantageous to fuse the facets into one seat. The tibial facet was bisected in the centre of the joint, and a radius was found to match the curve of the articulating dome. In the biological joint, the facets closely mesh to provide this; the radius of the dome was then used as a guide rail for a sweeping extrusion with the same geometry as the frontal plane groove for the talus. This preserves the contact points of the edges of the tibial dome and provides a matching meshing geometry shown in Fig. 25.

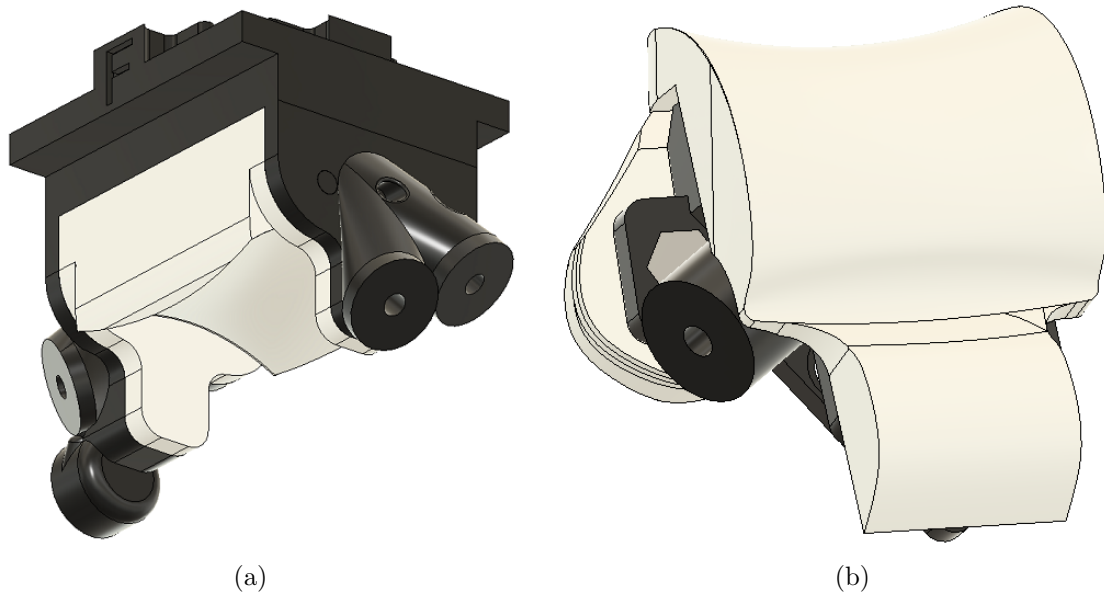


Figure 25: The isolated and simplified shank motion facet (a) in its mounting structure, and the talus superior simplified facet (b) attached to the centre scaffold.

The geometry of the inferior posterior facet of the talus is the most complex motion facet of the ankle. This motion facet is akin to an asymmetrical hyperbolic paraboloid. Similar to before, the facet was bisected in the centre parallel to the axis of motion. To

match the geometry, two fixed-point curves were connected. While predominantly flat, the anterior facet has a minor arc to the surface, which was matched similarly to the previous facets. The subtalar facets on the calcaneus were mirrored in those on the talus. The matching convex curve was formed for the posterior from the same sweep extrusion as that on the posterior of the talus. This facet was then brought to the anatomical position of the calcaneus and was oriented to match the angles of the local geometry. The cosponsoring process was then done for the anterior gliding surface, like its talar counterpart, which also has a minor but matching curve, and this was reproduced in the same manner. In addition to the motion facets on the calcaneus, anteriorly the bone increases high proximally, forming the structure for the motion facet with the cuboid bone [5]. This is similar to the notch on the superior talus, which forms a hard stop for the motion of the inferior talus. In the interest of stability, this structure was then formed from a cuboid and was then filleted to match the curvature of the top surfaces, shown in Fig. 26.

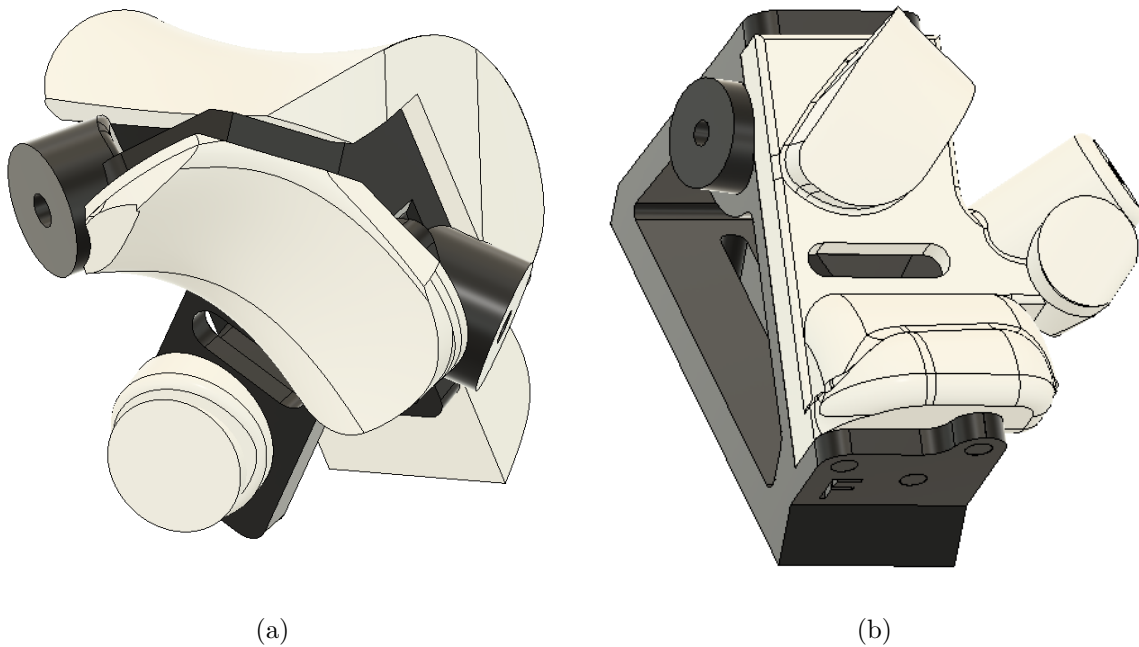


Figure 26: The isolated and simplified talus inferior (a) motion facets, and the isolated and simplified facets of the calcaneus (b). Also shown in the mounting structure representing the heel in (b) and the inferior view of the talus scaffold (a).

4.2 Ligament Design

4.2.1 Initial Shape and Specification

The second requirement for this project was to produce a simplified anatomical soft ligament with the capability of measuring its own deformation, shown in Fig. 27. A generic manufacturing format was designed to then differentiate into specific ligaments based on the anatomical measurements. Due to the small form factor of the joint, a standard 5 mm LED package, such as in Jamil et al, would cause an issue in effectively mounting the ligaments in the desired insertion points [19]. Both with designing an effective gripping method, transitioning to a mount and having the devices close together. To mount the ligaments in the insertion and origin sites effectively, bolts were used at both ends to directly attach to the *bone* via a hard eyelet. This required a non-sensorised section at either side of the operating region, which we will call the head of the ligament.

For the operating region of the ligament, the dimensions are largely dominated by the core size and the resulting cladding thickness. The core is largely dependent on the form factor of the photodiode and the LED required. The core was selected to be cast in Dragon Skin 15 by Smooth-On, as it had good transparency and its low viscosity meant it was easy to cast small objects. The available manufacturing process requires that the electronics be soldered by hand; therefore, parts cannot be too small for this to be feasible. The smallest IR LED and photodiode package found is the Vishay subminiature SMD devices TSML1020 and VEMD2000X01, with a form factor of 2.5 x 2 x 2.7 mm. This meant the core of the waveguide could be 2 x 2 mm; for this, a cladding thickness of 1.5 mm was originally chosen, leaving the thickness of the operating region to be 5 mm. The cladding was selected to be cast in Mould Star 15 by Smooth On as it was opaque and was matched to the same Shore hardness value of 15 as the core. While a Shore hardness of 15 is on the softer end of the scale, it provides greater flexibility to characterise the required strain of the ligaments, which can be reduced with silicones

with a higher Shore value. As for the ligaments themselves, 4 were chosen to be used in the design. Laterally, these were the calcaneofibular ligament (CFL) and the anterior talofibular ligament (ATFL). Medially, the posterior superficial tibiotalar ligament (sTTL) and the talocalcaneal ligament (TCL).

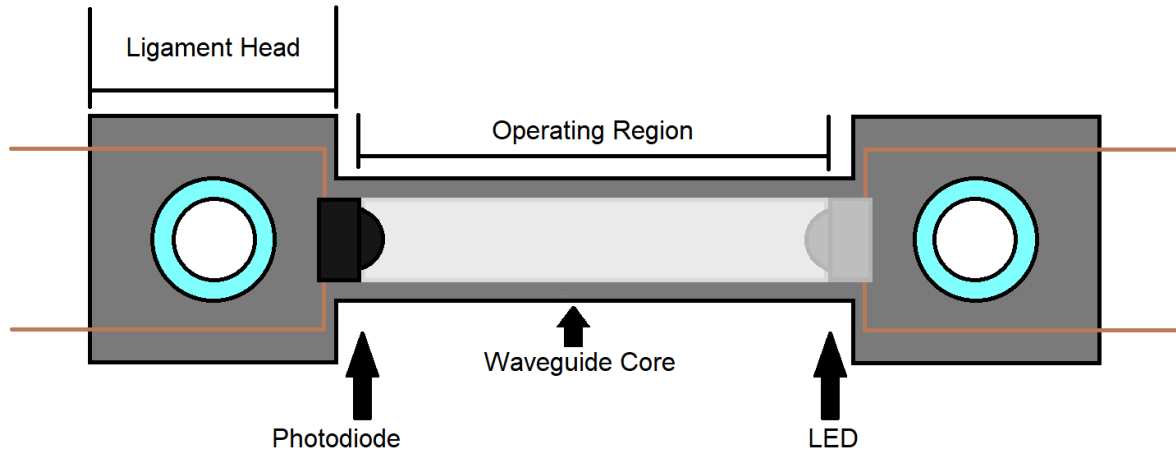


Figure 27: A cross-section of the ligament design.

4.2.2 Mould Design

The internal diameter of the eyelet size was chosen to have an M3 thread, and the outer diameter is 5 mm. The diameter of the head of the ligament was chosen as 10 mm, to allow 5 mm of bulk material around the eyelet for rigidity. To start with, a 2 mm thick wall was modelled surrounding the outer perimeter of the ligament shape, forming the mould wall. There was a concern that the head of the ligament would deform and come away from the eyelet during the operation, causing unpredictable behaviour. This was solved by using square-shaped heads, forming a *shoulder* shown in Fig. 27. This shoulder, alongside the material at the head, meant that a casing could be made around the head to restrict it from deforming, reducing eyelet separation. A two-part mould design allows for easier removal of the cast part. The base of the mould was modelled to be separate from the mould wall, shown in Fig. 28, so that the wall can be lifted from the base. The wall of the mould is extended at the bottom by 1 mm, matching a depression made in the surface of the base to locate the wall. Two grips are added to the top of the

wall for leverage during removal. In the base, two M3 holes allow a bolt to pass through from the top of the eyelet and are tightened with a nut on the underside.

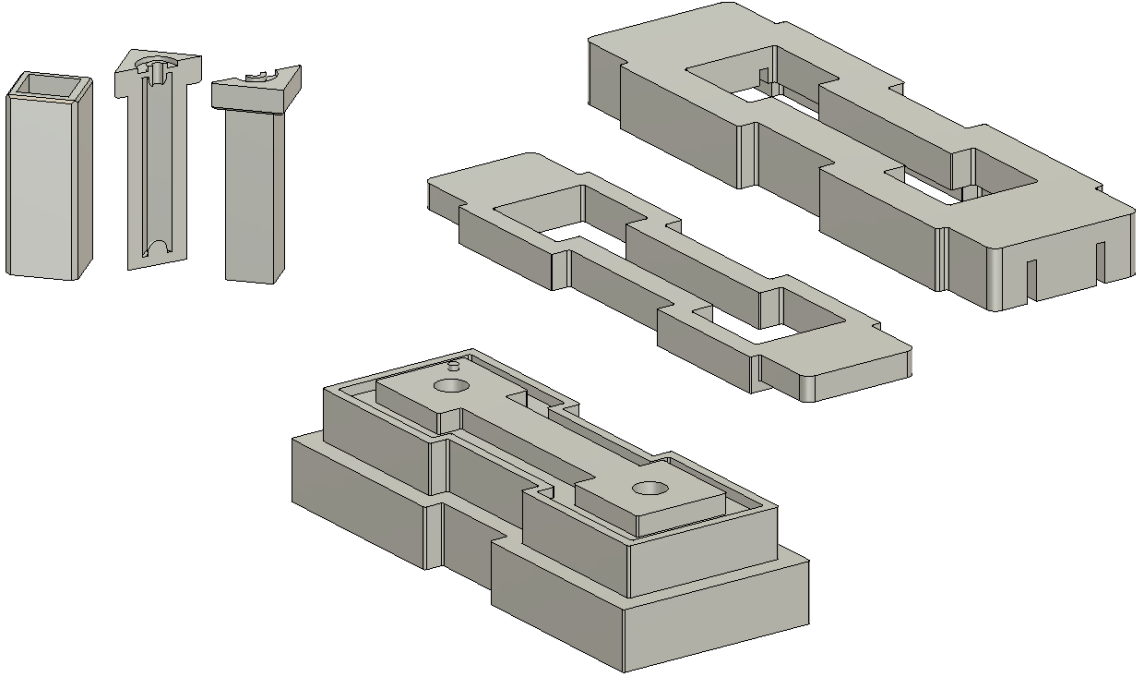


Figure 28: An exploded view of the mould design showing the removable wall, base, and eyelets. As well as the separate 2 part core mould d

In order to integrate a wave guide into the design, an optically transparent core, an LED, and a photodiode are cast inside the operating region of the ligament. To accommodate this, the mould was modified to accept a multi-material casting. Casting the core separately allowed for the opportunity to bond the electronics using an intermediary RTV silicone to the core before casting the cladding around it, preventing them from separating during operation. It is shown by Jamil et al that silicone is transparent to IR, which will allow external light to penetrate to the core and cause noise [19]. This was solved by pigmenting the silicone, adding a black pigment to the cladding during casting [19]. A black pigment (Smooth On, Silc Pig - Black) contains carbon black, which is highly absorbent to the IR spectra, making it ideal to pigment the cladding with to reduce external noise and internal power loss [15, 36].

High gauge copper wire is soldered directly onto the leads of the photodiode and the LED. A transparent core is cast, with hemispherical indents for the heads of the LED and photodiode, in a traditional style two-part injection mould from a syringe. The LED and photodiode are then adhered to the core using transparent RTV silicone (Dowsil, 732 Multi-Purpose Sealant) and left to cure. Once cured, the assembly is then suspended in the mould from the wires. The pigmented silicone mix can be poured into the mould, encapsulating the core as one piece. The head capsule was designed and printed with a wall thickness of 1 mm. The capsule had an increased thickness of 2 mm at the shoulder to hug the first 2 mm of the operating region and further reduce deformation. The interosseous talocalcaneal ligament was made by designing and printing a one-use breakaway mould in PLA. The iTCL will not be sensorised, as here it serves a purely mechanical role to prevent the subtalar joint from dislocating. Therefore, a one-part casting with a break-away mould was sufficient, as shown in Fig. 29. The iTCL was cast with a length of 11 mm with a width of 4 mm, with the insertion areas described in Section 4.1.5.

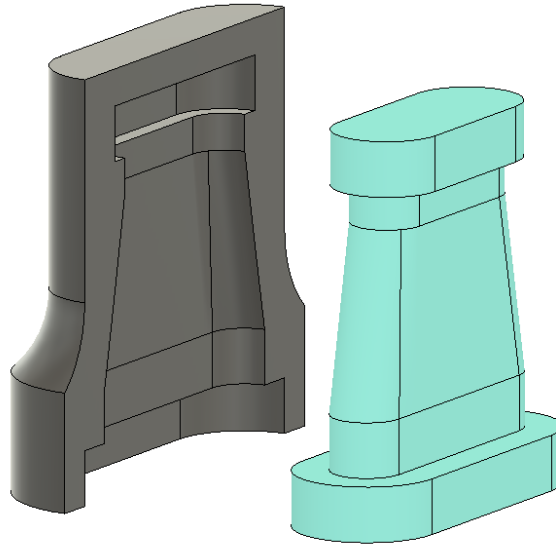


Figure 29: An image of the finished casting of the iTCL, demonstrating the pigment, head capsule and wiring.

The final step of manufacture was to differentiate the general mould design to the specific lengths of the 4 ligaments. The ligament lengths were chosen to be 75% of the maximum

of the ranges quoted in the literature review. This was chosen as an overestimate of the average, as the ankle design is unaware of the exact contact points due to the loss of cartilage information in a CT scan. The CFL was chosen to have a length of 31.47 mm. The TCL with a length of 27.85 mm, the ATFL with a length of 22 mm, and finally the sTTL with a length of 15.85 mm. The ATFL at 75% would have been 21 mm; this was tuned by trial and error to 22 mm. The ligament length is considered to be the maximum distance between the outer diameters of the eyelets. The linear deformation of the ligament is intended to occur in the region between the two heads, where the waveguide is embedded; this will be referred to as the operating region. The distance between the eyelets was changed to allow this distance to reflect the 4 specified lengths of the anatomical ligaments. For the shortest ligaments, the ATFL and the sTTL, the operating region was originally short, 0.85 mm for the case of the sTTL. Concerns of detector saturation due to proximity to the emitter and manufacturability led to the heads of each ligament being reduced by 1 mm on the operating side to allow for a longer operating region and therefore waveguide. The distance between the eyelets, with the anatomical distance of the ligaments, remained unchanged.

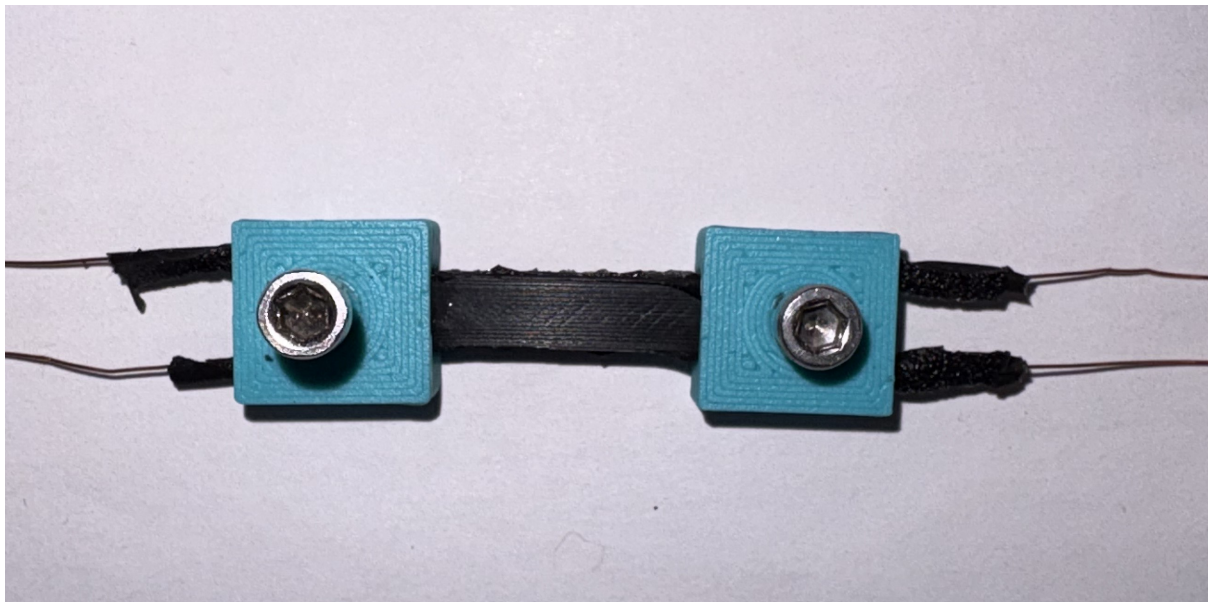


Figure 30: An image of the finished casting of the ATFL, demonstrating the pigment, head capsule and wiring.

4.3 Manipulator Rig Design

In order to manipulate the ankle joint to study its motion, a device must be designed to permit this consistently. The two key planes of motion in question are the sagittal plane and the coronal plane. A 2-axis gimbal will allow rotations in 2 planes orthogonal to each other. As the axis of rotation in each plane on the gimbal is typically a pin joint, the input rotation can be provided and measured here. Due to the translational addition to the rotation of a rolling contact joint, the centre of rotation will move during rotation. This means that a fixed shank relative to a fixed gimbal will cause unnatural movement and even binding in the joint surfaces. Therefore, the shank mount must be able to translate freely in the sagittal plane, as this is the major axis of movement for the ankle. A cube of aluminium extrusion provides a rigid square base, 20 mm x 20 mm extrusion was chosen, forming a 200 cube, shown in Fig. 31. The external gimbal ring and gimbal mounts were 3D printed in PLA. The platten was laser cut as a 120 mm disk from clear acrylics, with indents on two sides to attach mounts for the rotary axis. A needle from a 180-degree protractor is fixed to the platten mount at the rotation point orthogonal to the surface. On the outer surface of the gimbal ring, a flat mount is fixed to the ring, with the pin passing through it, and the dial of the protractor is fixed to this. This means that as the platten rotates, the dial moves with it, and the protractor dial can be used to read the rotational input from the perspective of the gimbal ring.

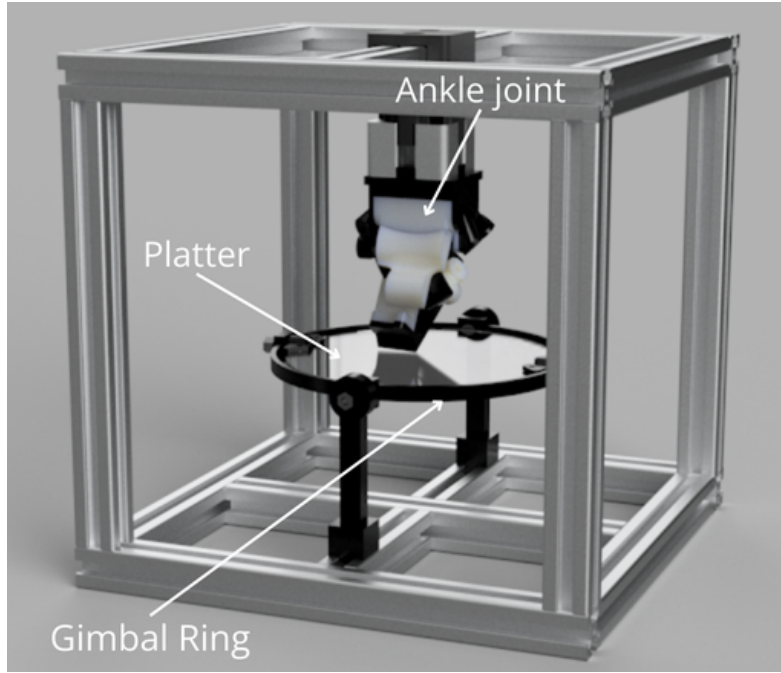


Figure 31: The manipulator rig design with the ankle joint mounted inside, showing the manipulation platter and the planar manipulation gimbal ring.

4.4 Ligament Testing

4.4.1 Tensile Testing and Linearity

After the waveguide embedded analogue ligament was designed, it was crucial to verify the functionality and performance of the device. The primary goal of these ligaments is to self measure its linear deformation, characterising the linearity was a necessity to assess future testing when in application. To test the ligament in a controlled environment, a tensile test machine (XFS1000, Testometric) was used. The protocol for the tensile tester was to extend and relax the ligament for 9 cycles, for which a cyclic profile was defined inside the tensile tester, shown in Fig. 32. As it is unknown how repeatable the result was, 9 repeats were chosen to give a large data set. The profile cycle started with an extension of 100 mm/min, followed by a 2 second rest, then a retraction of 100 mm/min, followed by another 2 second rest. The profile repeated the cycle until 9 cycles were complete. The maximum extension of the sample was dependent on the ligament tested. The extension of the ligament is determined by the integer in millimetres that was closest to a 15%

strain. This was chosen to be double the maximum strain demonstrated in the lateral ligaments by Renstrom et al, 6%, with an added margin [35]. For the CFL, this was a 5 mm extension, representing a 15.8% strain. The TCL was tested with an extension of 4 mm, a 14.4 % strain. The ATFL was tested with an extension of 3 mm, a 13.6% strain. And finally, the sTTL was extended to 2 mm, representing a strain of 12.6%.

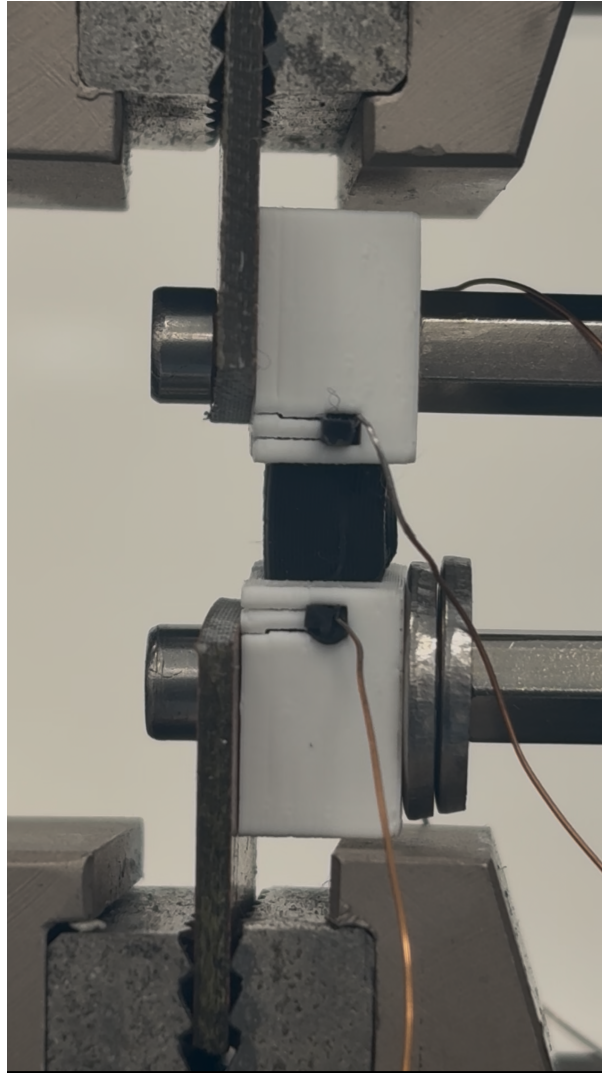


Figure 32: The test ligament installed in the jaws of the (XFS1000, Testometric) tensile test machine.

The waveguides are manufactured in silicone, which is a viscoelastic material. While viscoelastic properties tend to be temperature-dependent, silicone appears to be unique as it has excellent viscoelastic properties at room temperature [26]. As a precaution, the linear extension testing done to calibrate the ligaments was done in a temperature-controlled

room to mitigate small variances in properties within a test and for more accurate comparison between the 4 ligaments.

For data capture, the internal IR LED was powered from the supply of a development board (Mega 2560, Arduino), with a potentiometer to have fine control over the input voltage. The TSML1020 requires a minimum supply voltage of 1.2 V and a maximum of 1.5 V. The initial power at the detector is dependent on the emission of the LED, which is dependent on the input voltage. Fluctuations of input power for the waveguide emitter would introduce inconsistency during measurement. While the systems already account for variation in initial conditions with the conversion to power loss, this would drift if the input power is inconsistent. To help mitigate this, a bench power supply with a current limiter was used to reduce current spikes and fluctuations during measurement. Each test required the device to start with a measured voltage of 4.5 V, so it can be ensured that the measurement is not out of range, and any variation will not cause it to exceed the 5 V supply voltage. The potentiometer then both limits the supply voltage and allows the initial condition of the waveguide to be tuned as close to 4.5 V as possible before each test. The detector of the waveguide was connected to a transimpedance amplifier circuit, of which the output was measured by an internal ADC on the development board. The readings from the detector were sent over serial to a computer, where a custom Python script isolated detector voltage, generated and synchronised a time stamp for each voltage reading, then exported the whole test in CSV file format. Then the potentiometer is adjusted to bring the measured detector voltage as close to 4.5 V as possible. After which, the amplifier and surrounding circuitry are left to settle for 15 minutes before the test. Then the Python script is started to initialise data capture, and the cyclic test profile is started on the tensile tester. After the test profile has completed, the capture script is stopped, automatically exporting the CSV time series file, which is then imported to MATLAB for analysis.

Analysis in MATLAB began by converting the CSV file into a time series table in the workspace. For each trial, the start and end points of each cycle are manually identified, and each cycle is isolated. The derivative of each trial is used to determine the data point before it begins, and the voltage decreases. The same is done to find the data point after the voltage stops decreasing. The input voltage of the LED will never be truly constant; therefore, the output intensity will vary a small amount. This is especially significant when powering up or at the start of a cycle. Therefore, to decrease the uncertainty, we can express the signal from the ligaments as a function of power. By characterising the extension by a function of power loss, we remove the inconsistency in the voltage by tying it to a known reference point at the start of the extension, effectively measuring the power loss across the waveguide:

$$\text{Power Loss (dB)} = 20 \log_{10} \left(\frac{V_o}{V_{in}} \right) \quad (1)$$

As the raw data is a voltage time series, the time of extension must also be converted to the extension of the ligament. The strain rate of the test device was 100 mm/min, as the test is recorded in seconds the therefore:

$$\text{Extension (mm)} = \text{Time (s)} * \text{Strain Rate (mm/s)} \quad (2)$$

Where the strain rate is $\frac{5}{3}$ (mm/s). The extension of the ligament is then normalised to find the percentage strain:

$$\text{Strain} = \frac{\text{Extension (mm)}}{\text{Sample Length at Rest (mm)}} \quad (3)$$

After normalisation, the outlier cycles are removed, and an average of each of the cycles is taken, alongside the standard deviation, $\pm 1\sigma$. Then to assess the linearity of the device, linear regression was used. The MATLAB function **fitlm** was used to find a linear model of each average taken for a ligament. The linear object expressed by this function directly

assesses the similarity of the original plot to the linear model by means of a R^2 value.

4.4.2 Ligament Stretch During Ankle Manipulation

The characterised sensorised ligaments were tested in situ on the manufactured ankle joint to map their behaviour across the range of motion of the joint. This is the primary test of the study, which aims to measure the power loss at the detector of a ligament during the rotation of the joint in a single plane at a time. The plane that is not being tested will be placed and locked at rest, or 0° . The test structure is identical, no matter how many ligaments are being actively measured, to provide flexibility of testing, allowing for the study of single ligaments at a time if necessary. Before the test, all three structures of the ankle are prepared by wiping the contact surfaces clean, ready to load into the manipulation rig. The calcaneus is bolted to the platen of the rig via an adjustable slit on the sole and is adjusted to be perpendicular to the sagittal axis. Then any non-observed ligaments are bolted to the origin points on the shank, and the shank is slipped onto the linear axis on top of the rig. The talus was then held in place at the tibiotalar joint, and the assembly was aligned with the calcaneus. Then the non-observed ligaments are fixed to their respective origin and insertion points, and the heads of each ligament are aligned to have the maximum possible line of sight to each other.

Both the sagittal and the coronal axes are then set to 0° , and the observed ligaments are fixed on. Once completed, the source voltage is adjusted on each ligament so that the voltage at the detector is measured as close to 2.5 V as possible, as this will provide measurement travel in extension and relaxation. The protocol for measuring the motion planes is as follows. For the motion plane of interest, the initial detector voltages are measured at 0° , then the axis is rotated at 5° increments till the maximum rotation in the positive direction is achieved. During each increment, the non-observed plane is checked to still be at 0° , and the voltage of the detectors is recorded. Then the axis is brought back to rest at 0° , and the voltage is adjusted back to 2.5V to ensure each test is in-

dependent and not affected by the previous test, for a total of 3 repeats. The negative motion is tested in the same way, with 5° increments up to maximum, with 3 repeats. For the sagittal plane, the maximum motion in the positive direction, plantarflexion, is 35°, and the negative direction, dorsiflexion, is -20°. For the coronal plane, the maximum motion in the positive direction, eversion, is 10°, and the negative direction, inversion, is -15°.

The data processing for the manipulation test is similar to the routine for the tensile test procedure. However, the data is manually recorded during the test. As per the tensile testing analysis, the power loss of the ligaments is calculated using Eq. (1), but the x-axis remains in degrees for this observation. Each of the repeats was then averaged together to find the mean power loss at N ligaments for each of the measured planes of interest. The linear approximations of the ligaments found in the tensile test are then used to convert the power loss (dB) into percentage strain. As the rig is manually manipulated, and the angle is set by eye, it is important to understand the variance of each test. Therefore, alongside the mean, the standard deviation is calculated and is stored as $\pm 1\sigma$.

To gain an understanding of the ability to predict the angle of the joint from the ligament stretch of the ATFL and CFL, a multivariate linear model was produced to estimate the joint angle in the form of Eq. 4.

$$\theta = k_1\epsilon_{\text{ATFL}} + k_2\epsilon_{\text{CFL}} + k_3 \quad (4)$$

Here, θ is the observed angle of the joint, and ϵ_{ATFL} and ϵ_{CFL} are the measured ligament strains of the ATFL and CFL, respectively. The values of the coefficients of this linear model were solved using the matrix operation shown in Eq. 5, where the inverse of a 3 column matrix containing the values of the strain is multiplied by the column matrix of the angles.

$$\begin{bmatrix} \epsilon_{\text{ATFL}_1} & \epsilon_{\text{CFL}_1} & 1 \\ \epsilon_{\text{ATFL}_2} & \epsilon_{\text{CFL}_2} & 1 \\ \vdots & \vdots & \vdots \\ \epsilon_{\text{ATFL}_n} & \epsilon_{\text{CFL}_n} & 1 \end{bmatrix}^{-1} \begin{bmatrix} \theta_1 \\ \theta_2 \\ \vdots \\ \theta_n \end{bmatrix} = \begin{bmatrix} k_1 \\ k_2 \\ k_3 \end{bmatrix} \quad (5)$$

Multivariate error propagation is then used to determine the uncertainty in the angle estimation due to the variance in the two strain measurements. This is done by taking the partial derivatives of the linear model with respect to both ϵ_{ATFL} and ϵ_{CFL} , which are the coefficients k_1 and k_2 from Eq. 5 respectively. Then the standard deviation for the ATFL and CFL at each point can be used to calculate the uncertainty in the angle estimate, Eq. 6, due to the variance contributions of both strain sensors at that point.

$$\theta_{\text{uncertainty}} = \sqrt{(k_1 \cdot \sigma_{atfl})^2 + (k_2 \cdot \sigma_{cfl})^2} \quad (6)$$

5 Results

5.1 Tensile Testing

The linear approximations for each ligament tested are shown in Table 1, including the R^2 value. The averaged plots for all tested ligaments have been plotted for comparison with $\pm 1\sigma$, shown in Fig. 33.

Ligament Under Test	Linear Approximation	R^2 Value
Anterior Talofibular Ligament	$dB = -0.83666s + 0.087716$	0.999
Calcaneofibular Ligament	$dB = -0.826447s + 0.018726$	0.998
Tibiocalcaneal Ligament	$dB = -0.63393s + 0.029738$	1.000
Superficial Tibiotalar Ligament	$dB = -0.68445s + 0.16377$	0.996

Table 1: Tabulated data of the tensile test results, for the ATFL, CFL, TCL, sTTL.

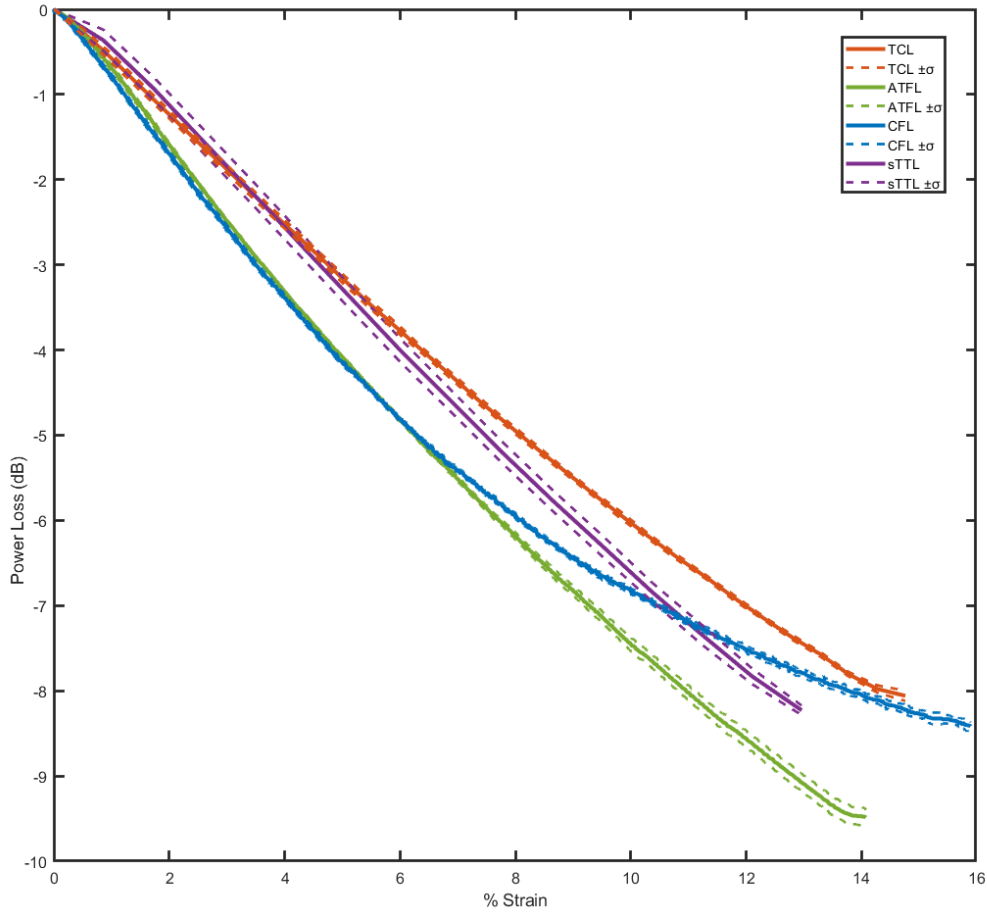


Figure 33: Power Loss (dB) of the TCL, ATFL, CFL, and the sTTL plotted against percentage strain (%).

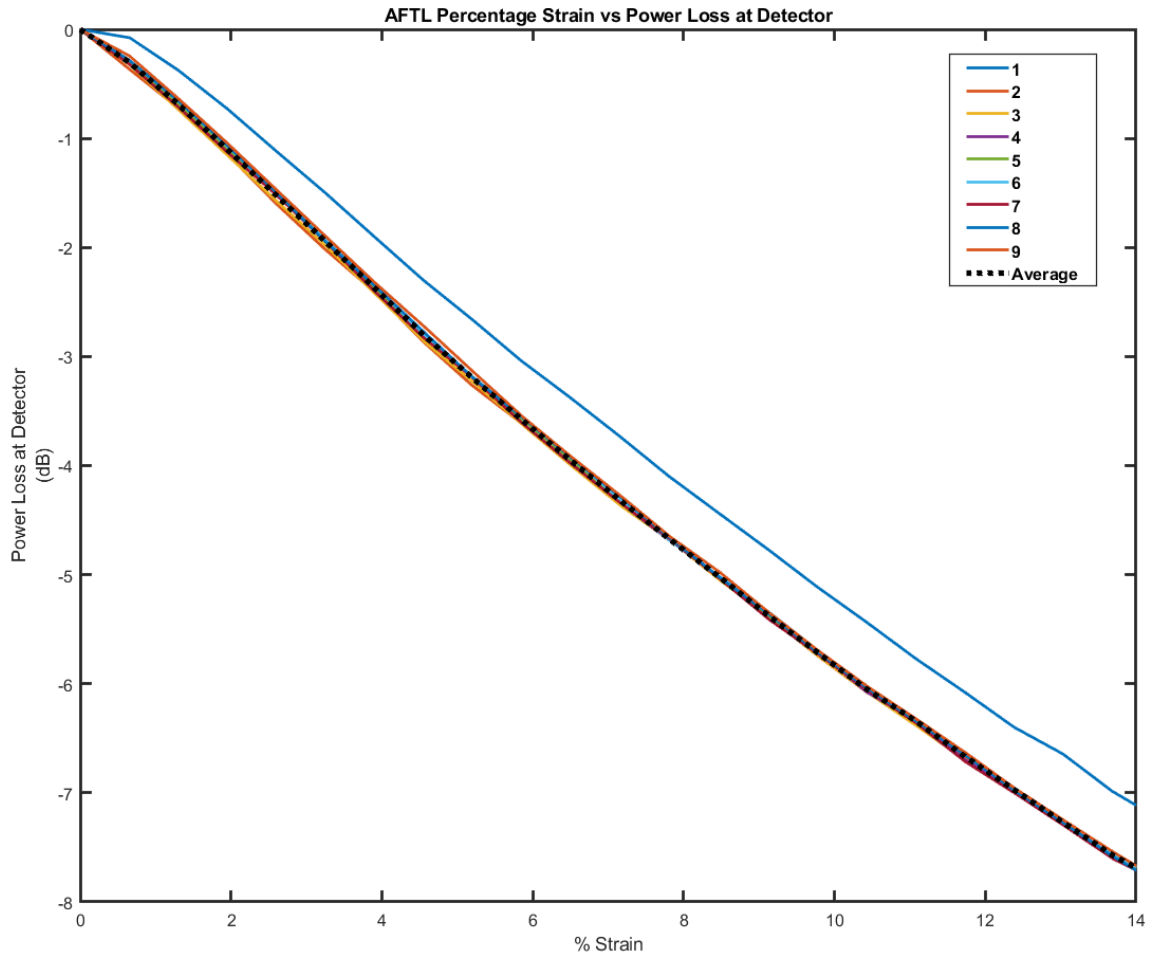


Figure 34: Power Loss (dB) of the individual and average ATFL runs plotted against percentage strain (%).

5.2 Motion Testing

The first primary test was the motion test conducted with the ATFL only in operation. All the other ligaments in the test were waveguide versions of the ligaments, but were powered off to reduce complexity in this initial test. Shown in Fig. 35 are the results of this initial test with the ATFL. A second test was conducted with only the lateral ligaments, the ATFL and the CFL, as shown in Fig. 36. The data from Renstrom et al has been plotted in Fig. 37 in the same style as the data of this study for comparison [35]

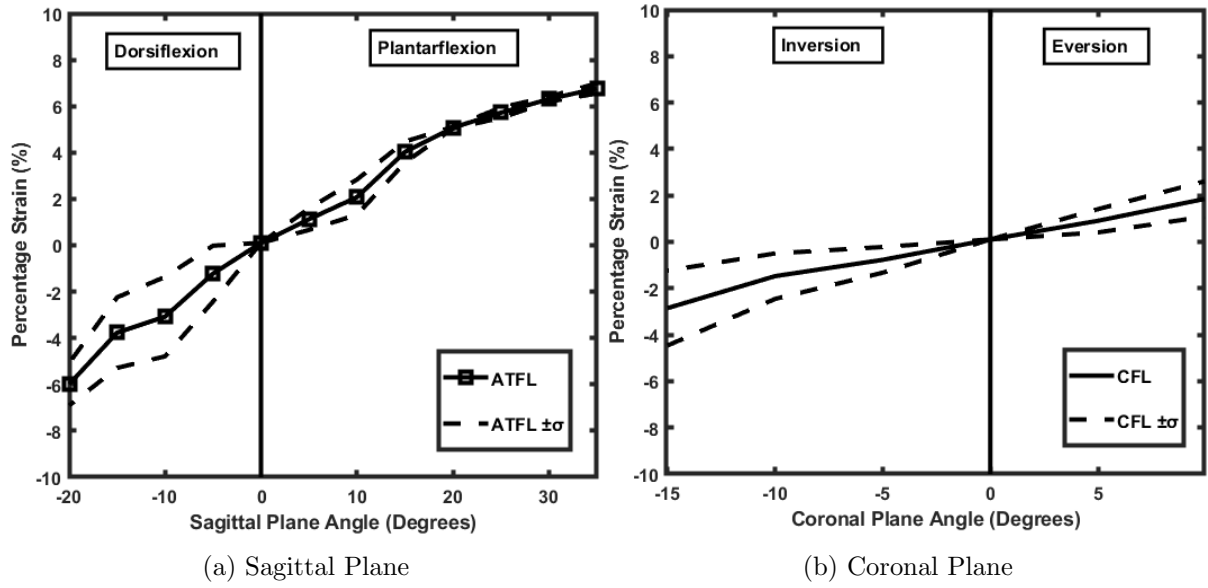


Figure 35: Power Loss (dB) plotted against plane angle of the ATFL for the (a) sagittal plane and the (b) coronal plane, respectively.

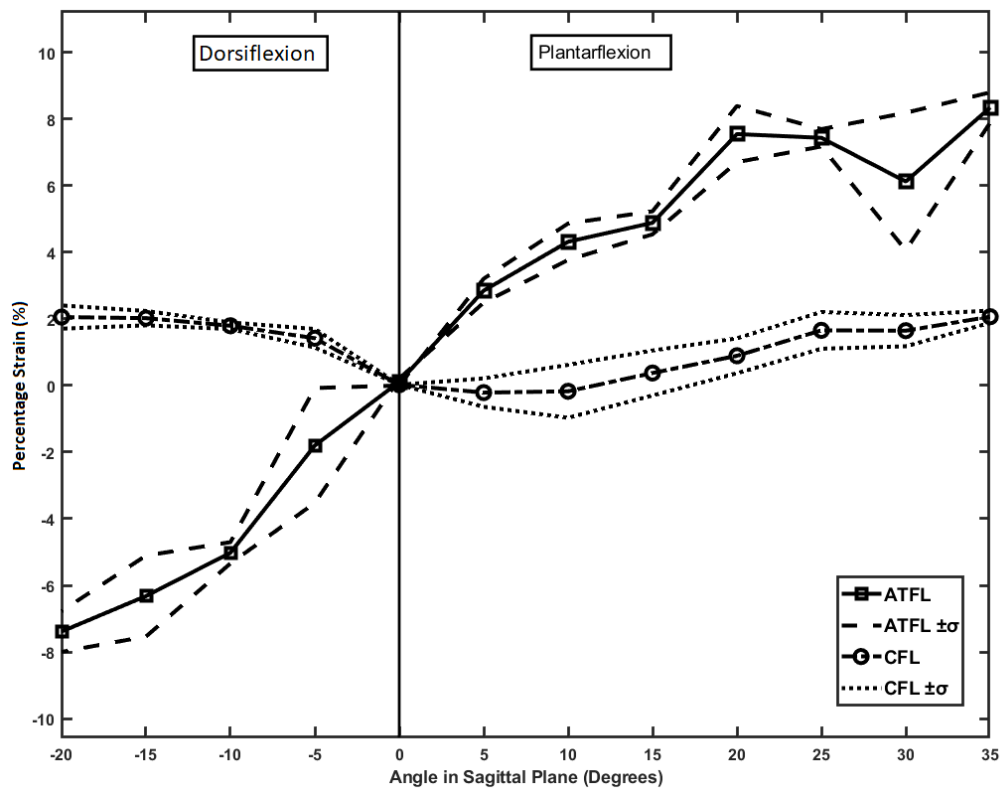


Figure 36: Percentage Strain plotted against plane angle for the sagittal plane of the ATFL and the CFL.

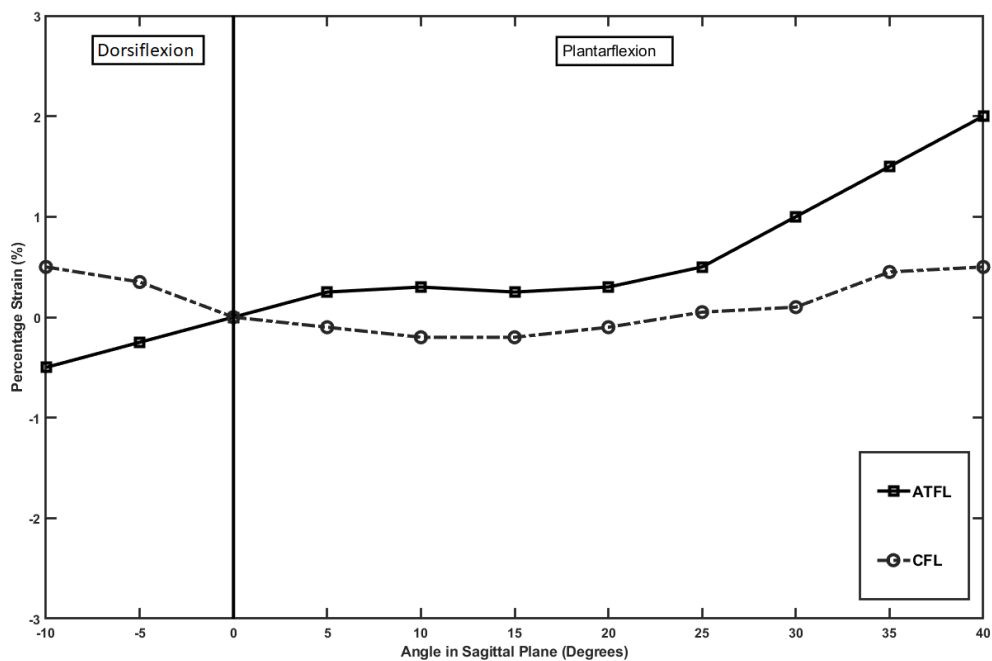


Figure 37: Percentage Strain of the anatomical ATFL and CFL plotted from Renstrom et al. Taken from [35].

The calculated linear model to estimate angle from the measured strain of both the ATFL and CFL was:

$$\theta = 2.162\epsilon_{\text{ATFL}} + 2.771\epsilon_{\text{CFL}} + 1.777$$

The uncertainty of the sagittal plane angle estimate, shown in Fig. 38, ranges from a minimum of approximately 0.3° to a maximum of approximately 5.8° , with the average uncertainty across the plot being 2.238° .

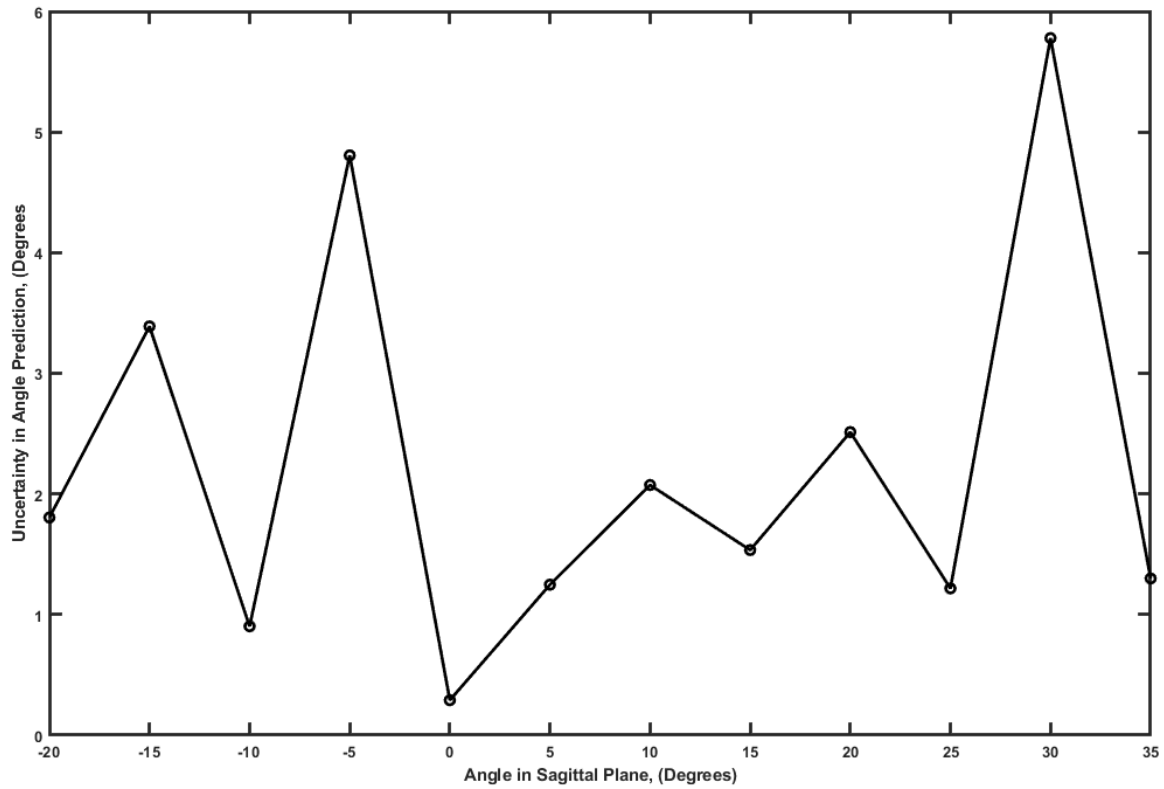


Figure 38: Uncertainty in degrees of the angle estimation due to the variance in the strain measurement of the ATFL and CFL.

6 Discussion

The design of the anatomically simplified rolling contact joint with proprioception has resulted in a multiplanar ankle joint complex consisting of the tibiotalar and subtalar joints. Each joint consists of the major articulating facets of the tibia, fibula, talus, and calcaneus, which have been isolated and their topologies geometrically simplified. Skeletal scaffolds were produced to preserve the anatomical positions of the isolated facets, which also serve as the mounting surface for a majority of the origin and insertion sites of ligaments. The analogue deformable waveguide embedded ligaments both constrain and provide feedback on their own operating strain.

6.1 Tensile Testing of Ligaments

Testing has shown that this design of the silicone deformable waveguide is capable of successfully measuring its own strain by the inference of power loss at the detector due to the increase in transmission distance of the light as it stretches. It was observed that the linearity of the sensor was capable of measuring the strain accurately within the operating range of 0% strain for every ligament tested. The conversion in Eq. 1 grounds the signal to a known voltage reference, mitigating any erroneous fluctuations in the optical power of the source. These could be purely noise, but also inconsistent effects from an uncontrolled operating environment, which will cause the initial and continuous optical power of the LED to never be identical across different tests. From the perspective of linear strain, they are capable of self-reporting their strain at an acceptable level of linearity.

The ATFL ligament was averaged across all 9 runs. Shown in Fig. 34 is the first cycle in the test sequence, which does not follow the average. Initial investigation with dry runs of this test showed that the first cycle consistently deviates from the trend of the remaining test. It was considered that while this "settling" in the initial run was marginally different in amplitude, the general trend held and was included in the av-

erage. The CFL ligament was also averaged across all 9 runs. Again, there is another period of "settling" within the first cycle of the test of the CFL. However, the first cycle has a different trend from the remaining 8 cycles. The first 8% follows the trend, with a slight amplitude deviation; this deviation increases for the remaining 8%. This deviation is less than that of the ATFL first cycle and is left to be included in the average. For the remaining 8 cycles, the trend here significantly differs from the 3 other tensile tests, with a clearly visible area of non-linearity starting at around 7% strain. For the TCL, in this case, the average was completed for the last 8 runs, excluding the 1st trial. The amplitude of the 1st trial varied significantly from that of the subsequent 8 trials, and was considered to be more than just the identified "settling" error. For this reason, it was excluded from the average; the remaining trials held a clear and consistent trend and amplitude as before. The sTTL was averaged with all of the 9 cycles in the test. However, the trend is distinct, but the amplitude of the different cycles has more deviation than of previous tests. For this reason, the first cycle "settling" is not identifiable, as if it were present, it is contained in the wider trend.

There are some interesting observations and caveats to address, such as the **settling** effect demonstrated in the first cycle of the tensile tests. The ligament body, excluding the electronics, is cast from a composite of silicone rubbers. As silicone is a plastic material, it is assumed that this "settling" is the initially amorphous polymer chains deforming into the operating direction for the first time, and then behaves consistently after [10]. Before testing, the CFL ligament had become slightly curved along the operating region. This meant that while at rest, it could produce a 5 V output signal; when stretched out in the tensile test machine, the voltage reduced to a maximum of approximately 2.5 V. It would appear that lower initial light intensity caused a non-linear trend after around 7% strain of the test. However, the entire strain range was able to be captured with the decreased start intensity. It is assumed that the non-linearity presented at the beginning of the exponential decay of the detected light, or the saturation of measurable light by

the elongation of the waveguide. The linearity of the sensor is preserved in the assumed operating region of 0 - 6% and is deemed an acceptable approximation. Similar to the CFL, there are non-linear tails at the end of the remaining 3 ligaments. These artefacts begin at 0.5% strain before the of each end of the test. This is most likely an artefact of the tensile test machine. The strain rate is constant during the duration of the test, and therefore, the plot is approximately linear. This is most likely an artefact of the acceleration and deceleration of the tensile test machine. This means the results are only accurate at low strain, the linear region shown in Fig. 33. However, in the scope of this study, the motion subsequent tests required static measurement; therefore, in this case, the linear approximations can be used.

An earlier design of the ligaments had the wires from the photodiode and the LED running perpendicularly out of the sides of the operating region. As discussed in section 4.2.1, the decision was made to run the wires parallel to the operating region out of the heads of the ligament. As only the ATFL and CFL were ultimately used during the motion testing, only these were fabricated with the new wiring; therefore, the sTTL and the TCL were tested with the perpendicular wiring. As this was done predominantly for durability reasons, it was not considered that it would affect the signal during operation. However, as shown in Fig. 33, there appears to be some grouping present between the old wiring scheme and the new wiring scheme. Irrespective of the CFL saturation curve, the linear section of the trend closely matches that of the ATFL, both using the parallel wiring path. The same is true with the sTTL and TCL; their linear trends also match closely, both using the perpendicular wiring path. It is to note that the sTTL and the TCL begin to deviate in trend soon after the ATFL and CFL; however, due to the saturation of the CFL, it is difficult to tell when this would have occurred in normal operation.

The objectives for this project stipulate that a soft anatomical ligament is produced with the ability to self-measure displacement. What has resulted are 4 silicone waveguides,

with anatomically correct lengths to match the ATFL, CFL, sTTL, and TCL. It has been demonstrated that the waveguides can provide a linear signal within the expected operating strain of 0-6%. Robust linear approximations have been calculated from the experimental data, and therefore, displacement can now be measured from the voltage signal. With this, they fit the specification for operation in the context of the idealised ankle joint, and have exceeded the requirements of the objective.

6.2 Motion Testing of the Ankle Joint

Motion testing of the ankle joint has shown that the entire system is able to closely mimic the motion of the native ankle joint. The feedback observed from the measured ligament stretch indicates that the system is capable of providing proprioceptive feedback of the motion of the joint, and could be capable of estimating the joint angle.

For the initial motion test solely with the ATFL, both the sagittal plane and the coronal plane present clear trends throughout the duration of the test, shown in Fig. 35. There is a linear behaviour in the stretch and compression of the ATFL in both planes. Compression occurs in dorsiflexion in the sagittal plane and inversion in the coronal plane. Stretch occurs in plantarflexion in the sagittal plane and eversion in the coronal plane. The key observation is that there is distinctly different behaviour when comparing the negative and positive angles in both motion planes. In the sagittal plane compression corresponds to the dorsiflexion, whereas extension corresponds to plantarflexion.

However, there is a clear caveat to this observation as the trends of both motion planes are the same. When the ATFL is in dorsiflexion, it is in compression; however, when the ATFL is in inversion, it is also in compression. When the ATFL is in plantarflexion, it is in extension; likewise, when the ATFL is in eversion, it is again in extension. This means that with only the ATFL, it can be said the ankle is in dorsiflexion and or inversion, or also in plantarflexion and or eversion. Therefore, we can infer information about

the planar position, but due to the similar trends, there is little differential between the planes, and the information is tied together. This was expected, as with the signal from a single ligament, the information available from a high level is simply whether it is in compression or extension. This is a static rig, and therefore, the likelihood of distinctive events from those of a kinematic and kinetic test is unknown. Therefore, it was anticipated that the information from multiple ligaments was key to differentiating the motion planes. As it was identified from Renstrom et al that the strain profile of the lateral ligaments, specifically the ATFL and CFL, is mutually exclusive [35]. Therefore, the strain profile of the CFL should oppose that of the ATFL if it matches the anatomical data [35].

It has been observed that the lateral ligaments are more innervated than the medial ligaments [34, 12]. This indicates a higher degree of neurological importance in the lateral ligaments, and therefore are potentially a greater part of the feedback mechanism for the ankle. This was also chosen as a direct comparison by the work conducted by Renstrom et al, which is fundamental to how closely the idealised joint and the ligaments both move and stretch [35]. As well as the proprioceptive ability of the ligaments to measure their stretch.

The second motion test, which included the ATFL and the CFL, presented the data in percentage strain, demonstrating experimentally that this is the case. During dorsiflexion, like before, there is compression of the ATFL. Contrastingly, for the CFL, there is an extension. Therefore, in this instance, if the CFL is strained and the ATFL is compressed, we can determine that the joint is in dorsiflexion. For plantarflexion, the trend of the CFL is similar to that of itself in dorsiflexion. The gradient of the strain curve is shallower for plantarflexion; however, as plantarflexion operates 15° further, they both reach around 2% strain.

Due to time constraints, the coronal plane test of the CFL & ATFL test was not com-

pleted. The omission of this test removes the ability to directly support validation of the dual planar position while simultaneously measuring the strain of both lateral ligaments in this instance. However, the single and dual ligament tests were identical in setup; the CFL sensor was inactive in the single test; therefore, the behaviour of the CFL in both tests is comparable. The single test can be considered paired movement, as it lacks the degrees of freedom to isolate sagittal plane motion from frontal plane motion. However, the two-ligament sensor test has 2 degrees of freedom and can differentiate plantarflexion and dorsiflexion. This observation indicates that the sagittal and coronal motion couple could be uncoupled, as the 2 degrees of freedom can quantify the sagittal component of the paired movement and therefore infer the coronal component.

To assess the ability of the ligament strain of the ATFL and CFL to predict the angle of the joint, the linear model in Eq. 4 was formed. To estimate the accuracy of the joint estimation of the uncertainty, Eq. 6 was used to calculate the uncertainty across the sagittal plane shown in Fig. 38. The average uncertainty of 2.238 ° is considered acceptable in this context; however, it is increased by peaks in the uncertainty, notably at 30°. This was towards the end of the test range, where the rig was reaching its limits, improving the stability of the rig here would likely reduce the uncertainty in these areas, and across the test.

There is one outstanding objective, which states to verify if the joint follows the motion of the human ankle. For the sagittal plane, the design is provided by the tibiotalar joint. The tested range of motion for dorsiflexion was 20°, in comparison to the range of 10-20° quoted in the literature [5]. For plantarflexion, the tested range was 35°, in comparison to the range of 40-55° also in the literature [5]. The test range in plantarflexion falls short of the quoted range by 5°. The joint is physically capable of achieving 40°; however, there is a limitation in the rig design that would mean the shank would derail.

For the coronal plane, the subtalar joint is the principal joint in the design. The quoted range of motion of the subtalar was 22.6°inversion and 12.5°everion [31]. During the test, the coronal range of motion was tested to 15°inversion and 10°everion. It is important to note that the range of motion of the subtalar is an estimate taken from the calcaneus angle. Due to the relationship of the hindfoot joint, no estimated range of motion has been found to the author's knowledge [31]. The tested angle of the joint falls short of the estimation in the literature. It would be possible for the ankle to physically reach these angles, but to not damage the ligaments, it was reduced. This is a limitation of the silicone; the maximum strain exceeds that of anatomical ligaments, but it can handle less force as a result.

Fig. 17 & Fig. 18 show the strain profile of the CFL and ATFL ligaments across dorsiflexion and plantarflexion [35]. In this particular plot, Renstrom et al are manipulating the foot in several motion planes to determine the behaviour of the ligament strain in various planar positions [35]. The "neutral" line presented on both plots is the strain of the ATFL and CFL under only dorsiflexion and plantarflexion, isolated in Fig. 37. Renstrom et al range the manipulation from -10°dorsiflexion and 40°plantarflexion, where the data presented in this project ranges from -20°and 35°, respectively [35]. For the ATFL, there is a linear type trend of compression in dorsiflexion and a linear type trend of extension during plantarflexion [35]. This is primarily the same trend for the strain measured from the waveguide in ATFL in this study. For the CFL data measured by Renstrom et al, there is a logarithmic extension during dorsiflexion[35]. During plantarflexion there is a shallow period of compression ending at 10°, followed by an extension of the CFL till the end of the test [35]. This again is primarily the same general trend observed and measured from the waveguide in the CFL of this study.

There are however some key differences in the amplitudes of both strain profiles. As previously discussed it is obvious there is a clear difference in the characteristic strain of

the silicone used in the project and human ligamentous tissues. Ligamentous tissue will microscopically fail at 4% strain, ending the linear region, macroscopically fail at 8%, and the maximum strain at break is around 10% strain [30]. The cladding material used in the waveguide ligaments (Mold Star 15, SMOOTH ON) has a maximum strain at break of 440% strain [42]. In Fig. 36, we can see the ATFL ranges from approximately -7% to 8% strain, in comparison to the strain range of -0.5% to 1% strain presented in the Renstrom et al Fig. 17 & 18. Due to the difference in strain profiles, the anatomical ligaments are far stiffer than the silicone analogue, therefore the analogue ligaments will present with a higher strain for a similar force. This is what is indicated in the comparison of the tested data in the project and the Renstrom et al data [35]. Again, for the CFL the Renstrom et al data show a range from 0.5% to 0.25% strain, in comparison to the measured 2% to 0.1% from the synthetic ligament [35]. The strain difference is less than in the ATFL in this case, however the discrepancy can also be easily explained by the different strain profiles of the human and synthetic tissue.

The segmented bone geometries are correlative to the cartilaginous joint surfaces, but a method of extrapolating the exact surface geometry of the meniscus from the bone is currently unknown [33]. Ultimately, simplifications done to the surface geometries will deviate from the true surface geometry; however, it does provide an approximate geometry of the meniscus from the relative bone geometries. The primary aim of simplifying the joint surfaces was to provide a geometry that is more manufacturable from quantifiable arcs and splines, and the functionality afforded by ease of manufacture was prioritised over absolute joint surface accuracy. Overall, however, with the joints appropriately matching the anatomical range of motion, and the lateral ligament data matching the same trend as the anatomical data presented by Renstrom et al, the designed ankle joint has successfully met the 3rd research objective of simplifying the anatomy of the ankle while preserving an anatomically accurate 2 DOF range of motion joint.

The planar position information derived from the ligament strain in Fig. 35 and Fig. 36 provides the potential architecture of a control scheme operating this joint. The characteristics representing the discrete planar positions have been identified and can be utilised in the position control of this joint. As revealed by the literature, the addition of the 2nd degree of freedom improves balance in passive ankles and power consumption and balance in powered ankles. Xiu et al shows the improvement of ankle motion with a rolling contact joint [50]. This design demonstrates the ability to have both of these core concepts in one design. The idealised joint structures provide similar motions to the human ankle joint, while providing 2 degrees of freedom. The strain data from the waveguide embedded ligaments gives proprioceptive feedback on the position of the joint. This is key to active or semi-active prosthesis, as it will enable control and stability of the joint via the actuators. Using both of the motion planes when heel strike is detected, it would allow the foot to infer the angles of the contact surface as a 2D plane, allowing for the potential of greater improved control, by detecting steep angles in both planes, and even uneven surfaces.

7 Conclusion and Future Work

7.1 Conclusion

In conclusion, the aim of this project was ultimately to design, fabricate, and test a multi-planar, anatomically simplified rolling contact ankle joint with proprioceptive ligaments. This joint matched the static kinematics of the human ankle joint, while the soft synthetic ligaments are able to measure their own deformation via integrated sensors. These ligaments also serve as the primary motion constraint to the design, as is the case in the human anatomy. The resulting design process produced simplified tibiotalar and subtalar joints, proving an anatomical-like motion in the sagittal and coronal planes. The CFL, ATFL, TCL, and sTTL were implemented as waveguide-embedded silicone ligaments, where designed to match the anatomical lengths of the chosen ligaments, also acting as the main mechanical constraints of both sub joints.

The analogue ligaments were extended to 15+% strain to characterise their linearity. All 4 ligaments operated linearly at least up to the assumed operating strain of 6%. The static motion of the ankle joint was tested in the sagittal and coronal planes. The output from the waveguides of the CFL and the ATFL across multiple points in sagittal and coronal planes and was converted to represent strain. It was analysed that there are identifiable unique patterns to determine the planar position of the joint complex. This strain data was also found to match the anatomical data presented by Renstrom et al, as well as the range of motion data presented in the literature. This system inherently produces proprioceptive feedback due to the ability to characterise the strain of the ligaments of the joint. This proprioception could be used to inform a control scheme for a powered prosthetic utilising this joint. The unique strain patterns of the ATFL and CFL could then allow the joint angle of the ankle to be inferred, in away that would replace the role of an encoder measuring joint angle in a traditional pin jointed ankle prothesis. This control scheme could be used to infer topological information about the contact surface

in real time and adapt the control of the joint, providing a pathway towards adaptive, biomimetic control strategies.

7.2 Future Work

Consideration of population-level geometric variance by using multiple scans or methods such as statistical shape modelling to form an average model was excluded from the scope due to the increase in workload. A single CT scan was selected to establish a proof of feasibility. To validate how different scans affect the validity of the design, further joints would have to be produced from different scans to provide a population average with regard to functionality. Further testing should be conducted to include measurement of both the lateral ligaments and potentially the medial ligaments, to validate the assumption that the isolation of the sagittal plane motion can collapse the paired motion from the single sensor test. This joint was designed under the initial assumption that it would be used purely in the context of this motion study, and therefore, considerations regarding weight bearing were not made. Further design and simplification would have to be made in order for this to become a weight-bearing or even a usable prosthetic joint. Beyond this, the application of weight to the joint would enable study on how the motion and, by extension, the proprioception vary with differing degrees of joint loading. The manipulation rig in this project was both manipulated manually and measured manually by eye with the protractors on the axis of rotation. This will have caused errors in both the angle set and the angle measured due to human error and parallax error in the protractor. This could be improved by driving the manipulation rig with actuators, which would improve both accurate positioning and measurement. This would also allow for some degree of continuous measurement, rather than static data points. The range of manipulation for plantarflexion should also be improved to rotate the joint to its anatomical maximum range of motion. Finally, further data analysis could be conducted on the ligament stretch to enable identification of the angle in plane rather than the discrete states currently identified.

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Appendices

A Expanded Work

A.1 Word Conference of Biomechanics 2025 Draft Abstract

Proprioceptive Feedback from a Complaint Biomimetic Ankle Joint

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1 Introduction

Commercial ankle prosthetics primarily use one degree of freedom (DOF). Prior research has demonstrated that systems with two DOF improve on user balance relative to one DOF devices [1]. However, these pin jointed designs still lack the functional behaviour of a healthy ankle, primarily because they do not provide sufficient compliance or the sensory feedback necessary for effective control. To address these limitations, we have developed an anatomically inspired ankle joint with motion constrained by soft multiplanar links analogous to ligaments.

2 Methods

An idealised version of the tibiotalar and subtalar joints was constructed to provide motion in the coronal and sagittal planes. Deformable waveguides were produced, representing the anterior talofibular and calcaneofibular ligaments, to provide stretch sensing. The waveguides were tested up to 10% strain, and the extension was approximated with linear regression. The joint was manipulated in dorsiflexion to 20° and plantarflexion to 35° in 5° increments. The resulting ligament stretch was recorded to establish the relationship between joint angle and ligament deformation.

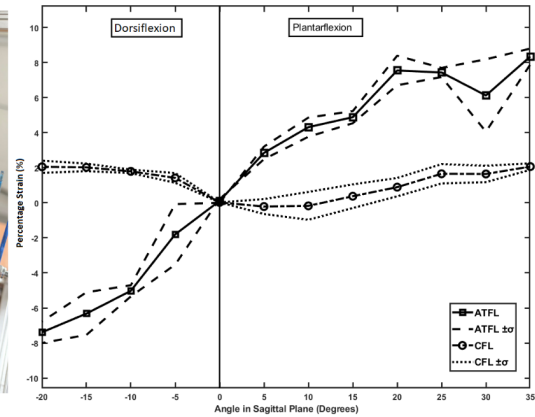
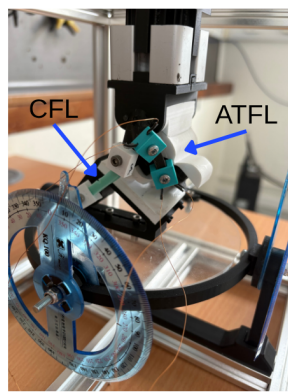
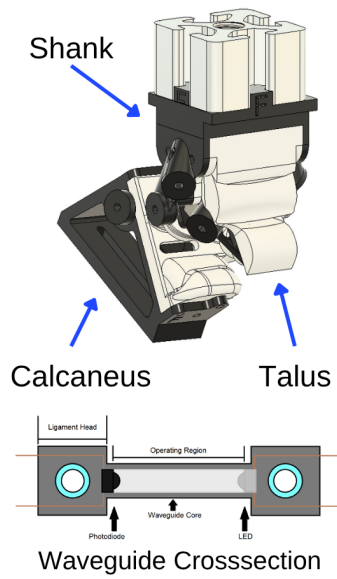
3 Results

The linear regression fit produced R^2 values exceeding 0.996 between the operating range of 0-6% strain. During joint manipulation, the measured stretch of the lateral ligaments exhibited a mutually exclusive pattern across discrete plantar direction of motion in both planes, as shown in Fig. 1.

4 Discussion

The measured strain behaviour closely resembles the anatomical ligament strain profiles reported by Renström et al [2]. This indicates that the simplified joint model can replicate physiologically relevant deformation patterns and that the ligament analogues provide directional information about joint motion. In addition, the use of soft, deformable ligament analogues inherently introduces compliance into the joint structure, offering a mechanical advantage over rigid pin-joint designs.

Because the system inherently produces proprioceptive feedback through ligament stretch, it could be used to inform a controller for a powered prosthetic joint. The distinct stretch patterns of the ligaments across motion planes may allow the controller to infer joint position, serving a similar functional role to joint angle measurements in conventional control schemes. When combined with the joint's compliant behaviour, this sensing capability provides a pathway toward more adaptive, biomimetic control strategies.



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