Physical sciences

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Abstract

In the original edition of Prosthetics and Orthotics International, Dr Sidney Fishman identified what he anticipated as foundational educational needs for the emerging field of clinical prosthetics and orthotics. Within the broader construct of the physical sciences, this included mathematics, physics, chemistry, biomechanics, and material sciences. The clinical application of these disciplines to expanding the collective understanding within the field is described, including the biomechanics of able-bodied and prosthetic gait, the material science of socket construction, the physics of suspension and load distribution, and the engineering of prosthetic components to mimic human biomechanics. Additional applications of the physical sciences to upper limb prosthetics and lower limb orthotics are also described. In contemplating the continued growth and maturation of the field in the years to come, mechatronics and statistics are suggested as future areas where clinical proficiency will be required.

Keywords

Physics, material science, biomechanics, mathematics, prosthetics

Introduction

In the inaugural issue of Prosthetics and Orthotics International (POI) in 1977, Sidney Fishman addressed what he anticipated as fundamental educational needs for the advancing field of prosthetics and orthotics (P&O). Within the physical sciences, this included mathematics, physics, chemistry, biomechanics, and material science. An enhanced understanding of these disciplines, he suggested, would be necessary to prepare practitioners in the field to raise the standards of a clinical professional. Thus prepared, these future practitioners would be viewed as co-equal members of the rehabilitation team, capable of providing individualized prosthetic and orthotic care, and prepared to contribute to the growth and advancement of the profession.

Using language that might surprise today’s P&O student, he began his introduction to the studies of physical sciences with a description of both the roles and shortcomings of a mechanic. “A mechanic,” he explained, may be defined as someone who performs manual work in the fabrication of some structure or device. The shortcoming in this approach is that the mechanic (or technician), not being exposed to many of the relevant principles evolving from the physical sciences, is taught to reapply the techniques he has learned in all situations with minimal variation.

If professionals in the field were to move beyond the reproduction of preconceived designs, they would need to better understand the complex interplay of the basic underlying sciences as the fundamental principles derived from such studies are requisite to the design and production of prosthetic-orthotic devices.

As suggested over 40 years ago, a working knowledge of the physical sciences has become foundational to the field’s ability to personalize solutions across a range of complex patient presentations and needs. The purpose of this article is to provide examples of how the physical sciences have informed advancements in patient care over the past 50 years and how such knowledge will need to expand in the next 50 years. This will primarily be accomplished through a consideration of the interplay between the physical sciences and lower limb prosthetic rehabilitation as chronicled within Prosthetics and Orthotics.
Hughes and Jacobs provided a definition of the gait: heel strike, foot flat, midstance, heel-off, and toe off, by the work of Perry, Winter, Gage, Sutherland, and locomotion would subsequently be changed and expanded. While this nomenclature and understanding of human locomotion would remain within reach of the clinical practitioner.

The narrative that follows will begin with kinesiology and its applications, first to our understanding of normal gait and then growing to include pathologic gait and gait deviations. Against this backdrop, the field could begin to contemplate socket designs, construction, and suspension methods. With an improved understanding of the human–device interface, the conversation continues with the evolution of components in general and feet in particular. In the realm of orthotics, once gait deviations are identified and understood, their remediation can be contemplated with respect to orthotic design and material properties. Throughout this process, the clinician is reminded that the hallmark of clinical care is to move beyond the repetitious actions of the mechanic into the personalized considerations of clinical care where the individual presentation of a given patient is considered and the physical sciences are used to optimize a solution.

As to the future, a working knowledge of the physical sciences will need to be supplemented with an understanding of the computer and statistical sciences. With the continued proliferation of microprocessors in P&O, the field will need to better understand the input and logic that informs the microcontrollers, as well as the mechanical response to such systems. Separately, a knowledge of mathematics will increasingly need to be supplemented with a working knowledge of the related field of statistics to ensure the findings of ongoing academic research remain within reach of the clinical practitioner.

**Gait analysis**

As if in response to Dr Fishman’s observations, within a few years of its inception, Prosthetics and Orthotics International published an instructive article on unimpaired human locomotion. Using the legacy nomenclature of heel strike, foot flat, midstance, heel-off, and toe off, Hughes and Jacobs provided a definition of the gait cycle, along with descriptions of the unimpaired kinematics and muscle activities of the hip, knee, and ankle. While this nomenclature and understanding of human locomotion would subsequently be changed and expanded by the work of Perry, Winter, Gage, Sutherland, and others, this early understanding of the physics of human motion, including inertia, external joint moments, lever arms, and measured kinematics allowed clinicians to begin to understand the deviations associated with prosthetic and orthotic gait.

Some 20 years after the work of Hughes and Jacobs, Sjodahl et al. published insightful work in Prosthetics and Orthotics International on transfemoral gait analysis before and after gait re-education. Applying the values of normalized gait speed, cadence, and the respective periods of single- and double-limb support relies on a foundation of biomechanics in general and gait analysis in particular. Similarly, in recognizing the power-deficit innate to unilateral transfemoral amputation, an understanding of the underlying biomechanics suggests the need to train toward increased ipsilateral hip flexion range, ipsilateral eccentric power in the hip flexors, and contralateral power in the ankle plantarflexors. The same background knowledge begins to explain why symmetrical transfemoral gait in terms of gait speed and step length may require reduced symmetry in pelvic motion.

Similarly, background knowledge in biomechanics facilitates an understanding of the adaptations required for running with a transfemoral prosthesis, the familiar “hop-skip” style, as described in Prosthetics and Orthotics International by Mensch and Ellis. The rationale behind the “hop-skip” running style recognizes the resultant ground reaction force that would be created at the prosthetic heel during traditional running, along with the resultant knee flexion moment and the inability of the ipsilateral hip to exert a sufficient extension moment to counteract prosthetic knee flexion. This thought process necessitates a foundation in biomechanics with an underlying understanding of dynamic forces and moments.

**The biomechanics of prosthetic gait**

With a foundation in gait analysis, the clinician is better prepared to understand the biomechanics associated with prosthetic gait across the various amputation levels. The inaugural publication of Prosthetics and Orthotics International featured such a publication on above-knee prostheses by one of the patriarchs of the field, CW Radeliffe. Drawing upon an obvious background in mechanical physics, this article introduces variations in polycentric and brake-type knee joint mechanics with additional considerations on their bench alignment. It introduces the concept of the “zone of stability” as it pertains to the maintenance of knee stability through the gait cycle, a concept that requires the clinician to understand the role of lever arm lengths as they act across various joint segments. The location and timing of force couples within the transfemoral socket are discussed, allowing the knowledgeable clinician to assess both the underlying mechanisms and potential solutions associated with transfemoral socket pressures.
Building upon this narrative, two years later, Foort described the biomechanical impacts of alterations to alignment of the transfemoral prosthesis such as overall lengthening, altered sagittal and coronal placement of the foot beneath the socket, and sagittal angulation (i.e. dorsiflexion and/or plantarflexion) of the foot. The relationships described by these variations with respect to socket pressures, step length, step width, and knee stability rely upon the readers understanding of transfemoral prosthetic gait and force couples within the transfemoral socket. This publication was quickly followed by Friberg’s study on users of transfemoral prostheses where it was found that, while roughly 80% felt that their prosthesis was the correct height, only 15% of the observed population were walking with a prosthesis within 1 cm of the correct length. While the majority of the subjects had excessively shortened prostheses, chronic low back, hip, and knee pain were significantly correlated with the lateral asymmetry caused by the incorrect length of the prosthesis, whether the prosthesis was too long or too short. A clinical understanding of the resultant biomechanics when a prosthesis is either too long or too short may have spared this population from considerable overuse injury and discomfort.

It was near this time that Hughes published a biomechanics article in Prosthetics and Orthotics International addressing the similarities and differences experienced by the user of a knee disarticulation prosthesis. The author draws upon the contrasting physics when a socket is loaded proximally, such as in a transfemoral prosthesis, and distally, such as in a knee disarticulation prosthesis, with respect to the resultant rotational effects. More specifically, with the proximal loading of a transfemoral prosthesis, rotational movements of the socket occur relative to the proximal brim of the socket, creating distal socket forces. By contrast, with the distal loading of a knee disarticulation prosthesis, rotation occurs relative to the distal aspect of the fully loaded limb, creating proximal movement and pressures at the brim of the prosthesis. The implications of such rotational forces were discussed.

As the field entered the 1990s, two more Prosthetics and Orthotics International publications by Schuch and Pritham explored the biomechanics underlying transfemoral socket design and alignment as the field transitioned from quadrilateral to ischial containment socket principles. In Schuch’s summary of an International Society for Prosthetics and Orthotics (ISPO) workshop on transfemoral fitting and alignment techniques, the UCLA contoured adducted trochanteric-controlled alignment method (CAT-CAM) prosthesis was introduced in detail, as were the alignment principles of Long’s line (defined as a straight line beginning at the center of a narrow medial-lateral socket brim, passing through the distal femur and distally through the center of the heel) and flexible inner sockets with supportive rigid frames. Relevant to this narrative, this material was followed by a discussion on biomechanics where it was suggested that, despite substantial differences in contours between ischial containment and quad socket designs, the underlying socket alignments and resultant biomechanics were largely unchanged. This position was revisited 2 years later by Pritham, who specifically addressed the underlying biomechanics, complete with pivot points and rotational lever arms associated with transfemoral socket alternatives. Specifically, Pritham addresses the now familiar need for the ipsilateral hip abductors to be placed in sufficient adduction to counter the moment created by gravity acting upon the body medial to the ipsilateral ischium. The resultant pressures along the lateral wall of the socket require a superior, medially-directed counter force, leading to the need for ischial containment. While the subsequent work of Gottschalk and Stills would suggest that the solution to the challenge at hand was more in the hands of the surgeons than the clinical prosthetists, the rationale behind the field’s transition toward ischial containment at that time was based on the underlying biomechanics acting upon the pelvis.

While the biomechanics of transtibial prostheses are largely addressed in the performance of their prosthetic feet, there has been some notable work in understanding the movement of the tibia within the residual limb, as well as the forces created and pressures distributed within the transtibial socket. While Radcliffe accurately used biomechanical principles and assumptions to describe the force couples acting within the transfemoral socket, 15 years later, Lilja et al. tracked the relative movement of the residual tibia within the limbs of seven subjects using X-rays to validate the biomechanical assumptions of swing phase pistoning of the limb within the socket and extreme anterior distal forces within the transtibial socket during heel contact. Five years later, Convery and Buis used force sensitive resistors to objectively measure the dynamic forces experienced within the patellar tendon bearing (PTB) socket. As theorized, their work confirmed high interface pressures at the patellar tendon bar and proximal posterior aspects of the socket with reduced interface pressures at the tibial tubercle and along the anterior tibial crest.

The year 2003 witnessed publication of two articles in Prosthetics and Orthotics International that described distinct concepts in transtibial socket design and pressure management with respect to positive and negative socket pressures. Kim et al. reported upon their observations in PTB-based targeted load bearing where they manipulated the depth of the patellar bar to determine the optimal depth with respect to subjective socket comfort. They found a patellar bar depth of 4 mm to be optimal among their participants. Meanwhile, the very next issue of Prosthetics and Orthotics International contained an article in which load-bearing pressures were managed through negative rather than positive pressure. In this study, Goswami et al. had patients with transtibial amputation walk in a series of sockets that were either at limb volume, 8% below limb volume, or 8% above limb volume. The authors...
found that the limb volumes of the subjects largely mirrored the available socket spaces, with larger sockets yielding progressively larger limb volumes. Notably, in the presence of elevated vacuum, larger socket volumes did not yield discomfort, pain, or redness.20 These studies are representative of the lessons clinicians were learning with regard to positive versus negative pressure applications in transtibial socket design.

With respect to the relationships between socket pressures and transtibial socket alignment, the field has recently begun to explore the concept of instrumented dynamic alignment, in which the measured reaction moments acting upon the socket could be used to inform necessary alignment changes. The pioneering knowledge required to inform such instrumented systems was published in 2016.21 In addition to tracking the flexion and extension moments measured in the sagittal plane and the valgus and varus moments measured in the coronal plane through the entire stance phase, this effort provided the mean and standard deviation values for these moments across 11 participants, suggesting that users associate a rather broad range of socket reaction moments with optimal alignment.21 The clinical value of such instrumented prostheses in obtaining ideal socket alignment has not yet been realized in routine practice, but remains an area of continuing exploration and study.

The topic of the biomechanics associated with amputation level would not be complete without mention of the creative contribution of Dillon and Barker22 to our understanding of the biomechanics of partial foot prostheses, published in Prosthetics and Orthotics International in 2006. This work challenged the prevailing assumption at the time that partial foot prostheses substituted the biomechanical functions of the forefoot by effectively restoring the foot length of the affected individual. This premise was examined by monitoring the anterior movement of center of pressure on the affected limbs of individuals ambulating with a range of partial foot prostheses. The authors noted that in those patients with more distal amputations fitted with toe fillers and slipper sockets, the center of pressure did not translate anterior to the distal aspect of the affected extremity until after contralateral heel strike. Translating the biomechanics into layman’s terms, the prostheses failed to extend the functional length of the foot. By contrast, in those with more proximal Chopart amputations managed with more traditional socket-style prostheses, the loads placed upon the rigid foot plate could be transferred to the anterior segment of the tibia, allowing the center of pressure to extend beyond the distal aspect of the limb prior to contralateral heel strike, effectively restoring the foot length of the residual limb.22

**Socket materials**

As the field became increasingly familiar with the dynamic forces acting upon the residual limb–socket interface, clinicians could continue to explore the range of materials available for socket construction. Over many years, readers of Prosthetics and Orthotics International have been introduced to a range of disparate socket materials including polymerized metal,23 carbon fiber,24,25 polypropylene,26,27 and nylon.28 Klasson’s24 publication on the theory and practice of carbon fiber in prosthetics and orthotics provides a comprehensive overview of this commonly used material. Here, the material science of carbon is treated at a foundational level, including the introduction of stress, strain, and the resultant modulus of elasticity; the concept of the I-beam as a mechanism for increasing the structural moment of inertia; the anisotropic behavior of carbon fibers as compared to the isotropic behavior of metals; fiber orientation; and permanent versus plastic deformation principles.24 Indeed, this is the type of paper that Dr Fishman would have likely welcomed into his 1977 curriculum.

This was followed 4 years later by a much more practical article, in which composite sockets were loaded to failure during a simulation of late stance phase.25 Of note, such studies of socket failures are limited by the absence of International Organization for Standardization (ISO) testing standards for custom-fabricated components. As a result, such studies are compelled to refer to the related ISO standards for non-custom prosthetic components. Unfortunately, among the combinations of the five different reinforcing materials and two resin types, none of the composites met the utilized ISO standards for level A100, with all failures occurring at the anterior aspect of the pyramid attachment plate through a range of shear, buckling, and tearing.25 This area of the socket was found to be the weak point of composite socket construction, with a need for carbon over fiberglass reinforcement.25

As the world continues to adopt three-dimensional (3D) printing in ever increasingly creative applications, one of the earliest publications on the use of selective laser sintering as a means of transtibial socket production was published in Prosthetics and Orthotics International in 2007.26 Describing the early experiences of the University of Texas Health Science Center at San Antonio, the authors recounted a feasibility study and subsequent evaluations of clinical acceptability. They also describe ongoing durability testing.26 Unfortunately, no subsequent publications on this topic from this team appear to have been published in Prosthetics and Orthotics International or elsewhere. However, strength testing of 3D printed sockets continues to mature.27–31

**Understanding the physics of suspension and hydrostatic load distribution**

Modern suspension options in lower limb prosthetic rehabilitation are based on an understanding of pressure gradients. The earth’s atmosphere exerts a moderate pressure of...
roughly 15 pounds per square inch on our environment. When the pressure of a closed system is maintained below that level, atmospheric pressure will exert a force upon that system. With respect to prosthetic suspension, there has been a progressive series of attempts to utilize this atmospheric pressure to push the prosthesis against the user’s residual limb by maintaining a low pressure environment within the limb–socket interface.

An early summary of manipulating pressure gradients to suspend a prosthesis was shared by Grevsten32 in 1978. Here, he observed that if a transfibial socket was constructed with additional distal space to receive downward stretched tissues via a pull sock through a distal one-way valve, the resultant pressure gradient would act to suspend the prosthesis. Citing earlier work, he reported a 1-cm reduction in vertical displacement or pistoning of the residual limb relative to the socket with the implementation of suction principles.32

Similar papers would follow with the evolution of various interface liners. Narita et al.33 observed a 1-cm reduction in vertical displacement of the residual limb compared to a cuff-suspended PTB prosthesis when the Icelandic roll on silicone socket (ICEROSS) system (Össur, Reykjavik, Iceland) combined the pressure gradient of their silicone liner with its mechanical locking system. Presumably due to the augmented tissue density within the liner, the same team reported an improvement in the dynamic angular stability of the residual tibia with the use of the ICEROSS system.33 Board et al.34 published their early work on elevated vacuum where an external vacuum unit was used to increase the pressure gradient between exterior environment and the sealed socket system, describing a 4-mm reduction in limb movement and a 7-mm reduction in residual tibia movement with vacuum assisted suspension relative to standard suction suspension. Brunelli et al.35 examined the ability of hypobaric seals to establish a proximal suction seal, comparing this approach to the more traditional sleeve-based suction suspension, and describing a 4-mm reduction during simulation of prosthetic swing phase. Collectively, these Prosthetics and Orthotics International publications have documented a host of approaches toward creating and sustaining the necessary pressure gradients to provide dynamic suspension of a prosthesis.

In contrast to the negative pressure gradients used to promote suspension of a prosthesis, in the realm of socket design and force distribution, the field has pursued the notion of hydrostatic loading of the limb through positive pressure gradients. By compressing the fluids of the limb, positive pressure is able to distribute body loads across the entire surface of the limb, thereby reducing localized areas of pressure.

Among the earlier publications on this concept was a Prosthetics and Orthotics International publication by Kristinsson36 in 1993. This served as an introduction to the mechanical concept of hydrostatic loading as an alternative to the legacy standard of the PTB socket. The now familiar concept of locking liners represents a combination of total surface bearing through hydrostatic loading of the limb in which the soft tissues are pre-compressed into a liner that mechanically locks into the socket and absorbs shear forces between the liner and the inner socket wall.36 The translation of these principles into the human experience was closely monitored in a series of papers where the benefits of silicone liners with regard to hydrostatic loading, limb cushioning, and comfort as well as improved axial suspension through associated pressure gradients and mechanical locking capacity were tempered with a recognition of their poor thermal conductivity.37–39 Similar reports were subsequently published on liners from alternate materials with different compressive properties, including both the gel-based Alpha liner (Willowood, Mt. Sterling, OH, USA)40 and the urethane-based testing of Total Environmental Control (TEC) liner (Ottobock, Duderstadt, Germany, formerly TEC Interface Systems, Waite Park, MN, USA).41 These were followed by a comprehensive examination of the thermal conductivity of various liner materials and thicknesses.42

**Mimicking human biomechanics through components**

Throughout its publication history, Prosthetics and Orthotics International has published a number of articles in which authors attempt to mechanically mimic human biomechanics through engineered solutions. Prominent among knee mechanisms was Radcliffe’s comprehensive analysis of the kinematics, alignment, and rationale behind four-bar linkage prosthetic knee mechanisms,43 and a precursor to the eventual stance flexion feature in the form of the so-called “bouncy knee,” described in 1985.44

An early approach to an axial torque absorber, the UC-BL Shank Axial Rotation Device from the University of California Biomechanics Laboratory, was described in the inaugural issue of Prosthetics and Orthotics International, complete with desired rotation range and spring torques.45 Specifically, these early authors suggested that such a unit would permit 20° of rotation in either direction with a centering torque of 0.23 Nm (2-inch pounds). These parameters were found to accommodate most daily activities without hitting the end stop of the unit and resulted in improved gait symmetry, reduced axial torques between the residual limb and the socket, reduced skin trauma at the proximal socket brim, and improved freedom of movement.45 While the evolution of such components received continued study and publication in other journals, it was just under 30 years later that a subsequent publication on this class of components occurred in Prosthetics and Orthotics International, measuring the peak torques and ranges of motion associated with both level-ground ambulation and turning with respect to both the inside and outside legs of the turning direction.46 Physiologic torques in this more recent analysis were much higher, reported at 8.2, 11.8, and 11.4 Nm on the outside leg during a turn, the inside leg during a turn, and during straight walking,
respectively.\textsuperscript{46} Approximately $20^\circ$–$25^\circ$ of rotation was observed, with most of that occurring in external rather than internal rotation.\textsuperscript{46} The observed torque absorption from commercially-available units varied with their installed elastomers but ultimately fell short with respect to their torsional stiffness.\textsuperscript{46}

With respect to feet, \textit{Prosthetics and Orthotics International} has chronicled many major transitions in technology. Shortly after its inception, \textit{Prosthetics and Orthotics International} described the ability of single axis feet to more closely mimic the biomechanics of normal ankle kinematics\textsuperscript{47} and temporal parameters of stance phase\textsuperscript{48} than the legacy solid ankle, cushioned heel (SACH) foot. Shortly, after the release of the Flexfoot (Ossur, Reykjavik, Iceland, formerly Flex-foot, Aliso Viejo, CA, USA) and Springlight (Springlight, Sandy, UT, USA) feet in the late 1980s, both of which used flexible keel designs, \textit{Prosthetics and Orthotics International} published an early examination on the material properties of stress, strain, and hysteresis in the presentation of the “Fle-skim,” a composite material made from fiberglass, carbon fiber, Kevlar, and methyl methacrylate resin that was used to provide an energy-storing shank between the transtibial socket and SACH foot.\textsuperscript{49}

As the field began to explore the concepts of energy storage and return, \textit{Prosthetics and Orthotics International} published one of the early articles examining the load deformation curves of numerous prosthetic feet, along with the resultant hysteresis loops, or the amount of energy lost during deformation and return.\textsuperscript{50} While weight and activity level are routinely considered in modern prosthetic rehabilitation when matching foot properties to their users, in this 1990 publication, the authors put forward the novel suggestion at that time that stiffness and hysteresis could be adapted to improve prosthetic performance.\textsuperscript{50} Their observations suggested that variations in materials led to variability in foot stiffness, and measurable effects of foot wear on measured stiffness values.\textsuperscript{50} This was followed several years later by a thoughtful paper on the mechanical properties of such feet and associated user experience.\textsuperscript{51}

Recent years have seen several publications on the results of various bench testing approaches to better understand the deflection properties and energy efficiency across a range of prosthetic feet, materials, and shoe types. Mason et al.\textsuperscript{52} evaluated a number of heavy-duty prosthetic feet against ISO 10328 standards and found that all tested feet passed these standards, but observed that the standards may be insufficient because they simulate only idealized gait. These same ISO standards were used in a subsequent \textit{Prosthetics and Orthotics International} publication that largely replicated the approach described in the 1990 study on load deflection properties and hysteresis loops using more modern prosthetic feet.\textsuperscript{53} This was one of the first publications to compare the dynamic elastic response characteristics of fiberglass composites against those of carbon fiber.\textsuperscript{53} Contemporary studies have recently examined the impact of the user’s choice of shoes and variations in coronal angulation and heel wedges, respectively, on measured energy storage performance values.\textsuperscript{54,55}

The rationale behind the field’s pursuit of feet capable, first of passive energy return and subsequently of active propulsion, can be explained in terms of the inverted pendulum model of human locomotion as articulated by Kuo and Donelan.\textsuperscript{56} Within this model, the stance leg is likened to an inverted pendulum with its fulcrum at the stance foot, with walking is seen as a sequence of efficient pendulum swings alternating from one stance limb to the other. Within this model, as the pendulum transitions to a new stance leg, a redirection of the body’s center of mass is required from the conclusion of one arc to the initiation of the next.\textsuperscript{56} A collision occurs with the leading limb striking the ground as the body’s forward movement is coupled with its downward descent from its greatest vertical height at midstance. Fortunately, the negative work experienced by the leading limb at this point in the gait cycle can be mitigated by the push-off of the trailing limb, shifting the body’s movement from forward and down, to forward and up.

The biomechanical value of the energy storage and return properties of prosthetic feet, described above, is a restoration of some degree of push-off to the trailing limb, ultimately mitigating the peak vertical forces acting upon the leading limb.\textsuperscript{57} Studies have suggested a direct relationship between the propulsive power of the trailing prosthetic limb and a reduction in the impact forces experienced by the leading, sound-side extremity.\textsuperscript{58} However, even with complete mechanical efficiency, the energy absorbed during the deflection of the prosthetic foot represents a fraction of the positive energy observed at the ankle at push-off. This deficit has led to the pursuit of powered prosthetic foot and ankle mechanisms capable of fully restoring physiological push-off.

Within the realm of replicating active plantarflexion during prosthetic push-off, an initial hydraulic-based concept and preliminary design was described in \textit{Prosthetics and Orthotics International} as early as 1985.\textsuperscript{59} This was followed 25 years later, by initial reports on a novel powered prosthesis in which pneumatic artificial muscles were rapidly inflated to successfully simulate the power of prosthetic push-off.\textsuperscript{60} In addition to these attempts to replicate the positive power of the trailing limb, efforts have been made to determine if variations in prosthetic design can reduce the magnitude of the negative work (i.e. impact) acting upon the leading limb. In one such study published in \textit{Prosthetics and Orthotics International}, investigators modified the longitudinal stiffness of unilateral transfibial prostheses by manipulating the springs within a shock-absorbing pylon across a broad range of longitudinal stiffness values.\textsuperscript{61} While the authors recorded both kinematic variables and kinetic forces, they ultimately concluded that...
variation in longitudinal stiffness did not appear to influence shock absorption for these prosthesis users.\textsuperscript{61}

A final construct related to prosthetic foot design and performance that has been described in \textit{Prosthetics and Orthotics International} is the novel biomechanical concept of symmetry in external work.\textsuperscript{61,62} This has been explored in a pilot case study to quantify gait differences between a diverse set of prosthetic feet\textsuperscript{61} and in a similar analysis in a larger case series several years later.\textsuperscript{62}

**The physical sciences in upper limb prosthetics**

While studies in upper limb prosthetics have been comparatively few relative to those in lower limb prosthetics, there have been a few examples within \textit{Prosthetics and Orthotics International} of the application of physical sciences to advance the clinical practice of upper limb prosthetics. Twenty-five years ago, Shaperman et al.\textsuperscript{63} measured the arm and shoulder strength of 37 children with unilateral transradial limb deficiencies between the ages of 3–5 years. Then, drawing upon prior publications, they examined the mechanical efficiencies associated with the activation of a body powered prosthesis equipped with both voluntary opening hooks and hands.\textsuperscript{63} This lead the authors to conclude that from a biomechanical standpoint, 3 to 5-year-old children were incapable of generating sufficient power to operate then-available voluntary opening hands.\textsuperscript{53}

Some 15 years later, a related study examined the mechanical efficiencies of available voluntary closing hand and hook prostheses.\textsuperscript{64} The authors' rather striking results found that the mechanical work required to generate a 15-N pinch force with a voluntary closing terminal device ranged from 33 N with an efficient hook design, to 131 N with the most mechanically-inefficient hand design.\textsuperscript{64} Examining the hysteresis of these mechanisms suggested that the energy dissipation of certain voluntary closing hands was 27 times greater than more efficient voluntary closing hooks.\textsuperscript{64}

**Applying material properties to orthoses**

The application of the physical sciences has not been confined to the field of prosthetics. Similar concepts have become integral to the practice of orthotics. Prevalent among these has been the manipulation of shape and construction to achieve desired biomechanical result with lower limb orthoses.

A foundational understanding of the underlying biomechanical principles was explored 20 years ago in a technical note that should be part of any entry level course on lower limb orthoses in which the author considered the resultant forces that occurred between the body and the ankle-foot orthosis (AFO) in the sagittal plane through early stance, late stance, and swing phase.\textsuperscript{65} Considering the underlying force vectors and biomechanics, McHugh points out that the forces are greatest when orthotic assistance is needed to compensate for plantarflexor insufficiency in late stance and comparatively small when the AFO is used to support the foot in the absence of dorsiflexion power in swing phase.\textsuperscript{65}

An early characterization of the relationship between AFO trimlines and their resultant mechanical behavior was described by Sumiya et al.\textsuperscript{66} In this study, the resistance to dorsiflexion and plantarflexion (i.e. mechanical stiffness) was measured on 30 AFOs as the trimlines around the ankle joint and posterior upright were progressively trimmed back, quantifying a greater maximum stiffness against plantarflexion (28 N·m) than that attainable for dorsiflexion (10 N·m) as well as a progressive reduction in stiffness with narrowed trimlines.\textsuperscript{66} Fifteen years later, this work was enhanced with a computer modeling effort in which the range of available stiffness values was calculated using variations in plastic thickness, the breadth and height of the posterior strut cutout, the radius of the posterior rectifications, and the transitional radii at both the superior and inferior edges of the posterior strut cutout.\textsuperscript{67} By manipulating these variables, the author calculated a broad range of potential stiffness values against plantarflexion between 0.04 and 1.8 Nm/degrees.\textsuperscript{67}

In addition to those efforts to understand stiffness and deformation in the plane of the applied loads, a related article reported upon the deformation of the AFO in all three anatomic planes when the device was loaded in a single plane.\textsuperscript{68} This confirmed the clinical realities of internal rotation of the calf section of an AFO when exposed to a dorsiflexion moment, as well as external rotation of the proximal AFO when plantarflexion moments occur.\textsuperscript{68} It was near this time that \textit{Prosthetics and Orthotics International} published a related analysis of the bending stiffness of six different colors of copolymer polypropylene.\textsuperscript{69} The authors described that pigmentation of copolymer altered its innate bending stiffness, but subsequently found that the variability in wall thickness inherent to drape forming appeared to negate the impact of such variability in the ultimate biomechanical performance of the device.\textsuperscript{69}

As the behavior of plastic was increasingly understood, the effect of both supplementary and replacement materials began to be explored. For example, Major et al.\textsuperscript{70} reported upon the resistance to dorsiflexion observed with four variations of the polypropylene AFO. Somewhat surprisingly, they reported that while the inclusion of L-shaped carbon inserts increased the resistance to dorsiflexion, it provided no more additional stiffness than the addition of an instep ankle strap that precluded the familiar “frogmouth” deformity often seen in terminal stance.\textsuperscript{70} In a similar effort, Sheehan and Figgens reported upon the observed stiffness and deflection observed during bench testing of off-the-shelf carbon AFOs (Orthotic Composites, London, UK) made to three stiffness values of Lite, Standard and Rigid.\textsuperscript{71} In addition to describing their approach to this novel
mechanical testing, the authors were able to quantify the differences in both tension stiffness and compression stiffness associated with the three material configurations and thicknesses.\(^7\)\(^1\)

Application of carbon reinforcement to knee ankle foot orthoses (KAFOs) has also been reported, in which the weight of the resultant device was 28% lighter than the standard plastic alternative.\(^7\)\(^2\) Additional examples of AFO material science are seen in early reports of AFOs produced through the additive manufacturing approach of selective laser sintering.\(^7\)\(^3\)

In contrast to prosthetic applications, where components simulate the biomechanical behaviors of various muscle groups during different phases of gait, the challenge in orthotics is often to supplement the actions of existing joints and muscle groups. An early example of this concept was found in the 1982 publication of Watanabe et al.\(^7\)\(^4\) in which plastic joints were put forward as a lightweight, noiseless, rust proof, and corrosion-free alternative to the more common metal joints. More recently, the addition of an oil-damper to an AFO was proposed as a means of replicating the eccentric contraction of the dorsiflexors in loading response when these are insufficient.\(^7\)\(^5\)

In a separate effort, the mechanical properties of Tamarack (Tamarack Habilitation Technologies, St. Paul, MN, USA) dorsiflexion assist flexure joints where measured as their locations were transposed anterior, posterior, superior, and inferior to their correct anatomical position.\(^7\)\(^6\) Anterior and posterior alignments were found to be particularly deleterious to the desired biomechanical performance of this joint in assisting with ankle dorsiflexion.\(^7\)\(^6\)

What may lie ahead?

_Prosthetics and Orthotics International_ has helped chronicle the development and broad application of microprocessor-regulated prosthetic technologies, from early studies of the intelligent prosthesis with microprocessor-regulated swing control\(^7\)\(^7\) to studies of microprocessor-regulated swing and stance control with the C-leg (Ottobock, Duderstadt, Germany)\(^7\)\(^8\) and Adaptive (Blatchford Group, Basingstoke, Hampshire, UK) knees.\(^7\)\(^9\) It published one of the earliest systematic reviews of microprocessor knees,\(^8\)\(^0\) as well as studies suggesting the benefits of stance phase microprocessor controlled knees to patients with limited community ambulation capacity.\(^8\)\(^1\),\(^8\)\(^2\) These publications from the last 20 years suggest that, in addition to mathematics, chemistry, material science, physics, and biomechanics, clinicians of the modern era will require a basic understanding of mechatronics, complete with an understanding of the types of electric sensors that might be used to inform the real-time logic of a microprocessor, to the logic and underlying programming of various microprocessor-regulated components, to the mechanisms of producing the mechanical adaptations triggered by sensor inputs. As the field continues its adoption of microprocessor-regulated prostheses and orthoses, it will be increasingly important for clinicians to understand how these complex algorithms function so as to best use that information when optimizing the device for a particular patient.

In his 1977 article, Dr Fishman included mathematics in his recommended studies of the physical sciences. While basic mathematics facilitates an understanding of physics and engineering, the complexities of modern publications now require an increased understanding of the related field of statistics. Thus, as modern publications continue to report on ever more complex statistical principles and methods, the clinical consumer of this research will need an adequate knowledge of statistics. As these statistical approaches increase in complexity, there is a genuine risk that most clinical practitioners will be unable to place the ultimate findings of such studies into some sort of clinical context. Thus, a background in statistics will become increasingly germane to the core competencies of the practicing clinician. Meanwhile, journals like _Prosthetics and Orthotics International_ will need to ensure that the nuances of academic rigor, the minutia and manipulations of gait-lab-derived biomechanical data, and the increasingly complex statistical methods and results do not ultimately alienate practicing clinicians who should remain the core constituency of the journal. Rather, efforts must be made to ensure that the messages of future publications are reasonably translated for integration into clinical care.

Conclusion

Dr Fishman’s prescience in anticipating the foundational educational needs of the field have proved to be quite striking. Within the physical sciences, the topics of physics, biomechanics, and materials science have become essential elements of understanding the field itself and enabling the clinician to tune their solutions to the individual needs of their patients. As the field continues to mature, the additional disciplines of mechatronics and statistics will become increasingly essential elements to individual patient care and integrating academic findings into the clinical environment.

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